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ABSTRACT

These proceedings represent a cross-section of the results of rehabilitation research across a range of disciplines. The 130 presentations, which deal generally with the exploitation of science and technology in the rehabilitative process, are divided among 21 sessions. Topics of the first 15 sessions are: aids to promote independent vocation and life activities, biomechanics/kinematics, improvements in wheeled mobility (papers with a clinical orientation), computer and microcomputer use, improvements in upper limb prosthetics, wheelchair design research, technology transfer, research in electrical stimulation (papers with a clinical orientation), studies in augmentative communication (papers with a clinical orientation), investigations into service delivery, functional assessment (much of it computer-assisted), rural rehabilitation, research on sensory aids, studies dealing with various aspects of lower limb orthotics and prosthetics, and more electrical stimulation research. The other six sections address these concerns: research into and aids to promote independent vocational and life activities; studies of control interfaces, including microprocessor systems; studies of biomechanics/kinematics; development and research in the area of alternative communication; development and research in wheelchair seating; and study of microprocessor system use. Twenty-two presentations from two poster sessions are also provided. Two student design papers on posture safety seats and myoelectric prosthesis control using microprocessors precede abstracts of 24 self-contained audiovisual presentations, the majority of which address clinical issues. An author index is attached. (YLB)

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PROCEEDINGS OF THE EIGHTH ANNUAL CONFERENCE ON REHABILITATION TECHNOLOGY

Technology-A Bridge To Independence

Memphis, Tennessee • June 24-28th, 1985

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RESNA Rehabilitation Engineering Society of North America
An Interdisciplinary Society for the Advancement of Rehabilitation through Technology

PROCEEDINGS OF THE EIGHTH ANNUAL CONFERENCE ON REHABILITATION TECHNOLOGY

Technology-A Bridge To Independence

Memphis, Tennessee • June 24-28th, 1985

Cliff Brubaker, Ph.D.
Editor

Douglas A. Hobson, P.E.
Conference Chair

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Conference Chair

The Rehabilitation Engineering Society of North America (RESNA) is an organization concerned with the exploitation of science and technology in the rehabilitative process. Its membership includes rehabilitation professionals (drawn from all pertinent disciplines), providers, and consumers. A goal of the organization is to promote interaction between these groups so they can better understand and serve the needs of those who can benefit directly from the application of rehabilitation engineering technology.

This volume represents an excellent cross section of the results of rehabilitation research during the past year. Its interdisciplinary character epitomizes the philosophy of RESNA; that is, that rehabilitation engineering is a team approach that requires the skills of many professionals--physicians, engineers, therapists, nurses, to name a few--as well as consumers, without whose input the advancement of rehabilitation technology would be severely curtailed.

This year's theme, Technology--A Bridge to Independence, is embodied not only in the scientific program that comprises these proceedings, but also throughout the instructional courses, exhibition and product demonstrations, and student design projects. Emphasis has been placed on the application of technology to encouraging presentation of clinical as well as research papers and posters. This year an audiovisual program was initiated which stimulated submissions of more than twenty quality presentations; the majority of which address clinical issues. Poster presentations have also become a forum for presentation at the RESNA conference. Twenty-one poster presentations were accepted and the papers are included in these proceedings.

Once again, we were fortunate enough to have a most enthusiastic and dedicated planning committee. In particular, we acknowledge the leadership of Cliff Brubaker, Ph.D., chairman of the committee that resulted in this record of our annual conference. We are also most indebted to the authors that submitted the quality material, and to our editorial and office staff that prepared this publication.

Traditionally, our annual conference proceedings embodies the most comprehensive and current statement on the state-of-the-art in rehabilitation technology. We believe this 1985 edition of the RESNA conference proceedings is also an outstanding contribution to this fine RESNA tradition.

Clifford Brubaker, Ph.D.
Chair, Papers, Posters,
Audiovisual and Special
Interest Group Program

Perhaps Memphis and 1985 will be remembered as the place and year of more positive additions to the RESNA conference program. Several noteworthy developments are included for the 1985 Conference. Authors were given the option of presenting papers at "Scientific" or "Clinical" sessions. While the logistics of scheduling made it necessary to combine papers in a few instances, the clinical papers are identified in the program with a (C) to aid conference attendees in their session selections. Self-contained audiovisual presentations are now an option for authors. These presentations are scheduled on a rotating basis during the conference and the abstracts of the audiovisual presentations have been added to these proceedings.

Probably the most significant new undertaking is the prospective formation of Special Interest Groups or SIGs, as they have come to be known among the conference committee.

It is difficult to prejudge the overall quality of contributions, but I will predict that the papers, posters, and audiovisual presentations accepted for the 1985 conference will be better on the whole as a result of more rigorous review standards adopted by the conference committee.

The tasks of organizing, reviewing, and scheduling the papers, posters, and audiovisual program fell on several sets of shoulders. I am indebted to the twelve chairpersons who organized their respective content areas and conducted the review process. They have also accepted the associated responsibility of chairing the Special Interest Groups. I would particularly like to express my gratitude to Sam McFarland for his work in the development of the program and scheduling, Olunwa Nwaobi for developing the audiovisual program, and Douglas Hobson for guidance and timely advice. It is obvious to anyone who has participated in a program of this nature that organization and preparation of the proceedings is best done by the professional staff. I would like to thank Patricia Horner and Susan Leone of the RESNA office for their efforts in compiling these proceedings.

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A MODEL OF COMPUTER AIDED WORK SUPERVISION FOR PERSONS WITH SEVERE LEARNING HANDICAPS

Robert A. Chubo
University of South Carolina

ABSTRACT

Available technology provides a means of replacing human intensive training and supervision required by persons having severe learning handicaps with a more economically viable computer aided model. A model system is described for supervising a parts sorting work task. Ethical issues and caveats are discussed.

BACKGROUND

Persons with disabilities resulting in severe learning limitations constitute one of the largest handicapped populations. Included are persons who are classified as being mentally retarded, as well as those with cognitive disabilities due to head injury, stroke, etc. Persons with conditions resulting in significant learning limitations pose an especially difficult challenge to rehabilitationists attempting to find a place in the work force for them. The learning limitations militate against the cost-effective acquisition of knowledge and skills essential to employment. "Cost-effective" is emphasized because it has been demonstrated that with intensive, structured training and supervision, many of these persons can acquire the skills necessary to perform relatively complex work tasks (1, 2).

However, a dilemma lies in the fact that the intensive training and supervision necessary for such accomplishments are usually considered too costly for widespread acceptance in the regular work place when compared to that required by nonhandicapped workers. Consequently, implementation of this programming has been largely restricted to sheltered, subsidized settings, such as workshops. Thus, there is need for a more efficient model of worker training and supervision for use with the severely learning handicapped population. Available technology may hold the key to resolving this dilemma. Technology can be used to reduce the human training/supervision requirement by replacing it with a more economically viable computer aided supervision system.

MODEL

A model of computer aided supervision has already evolved in the information processing business. In these systems, the word or other information processing programs have incorporated into them a monitoring algorithm which provides supervisory personnel with error rate, work speed and other productivity indicators. These computer activities are quite elementary and analogues are widespread. They can be found in most computer based programmed instruction and games. What is being proposed here is an extension of such a computer based monitoring system to accomodate the need of persons with severe learning handicaps, thereby enabling them to function more effectively in the work place.

To reiterate, all components of this system are currently available. They are used in computer based instruction, games, environmental monitoring and control systems, and robotics. These components can be integrated into work task instruction and monitoring systems which have the capability of performing many of the supervisory functions which are now human intensive and therefore, extremely costly. For example, a computer work task instructional system can be configured to provide repeated step by step prompting to the worker. The instruction can be presented in concrete forms utilizing computer graphics or videotape recordings of task sequences. These visual instructions can be augmented with aural, synthesized speech and video displayed text when warranted. Since the computer is tireless, if necessary, such step by step instruction can be provided on a continuous basis. In this very basic form, immediate utility is apparent for persons with brain dysfunction rendering them unable to commit new information to longterm memory.

A more elaborate and widely applicable system is feasable, however. An expanded system can include actual work task monitoring which is similar in principle to that used in information processing activities. If data from work task performance is input into a computer, several computer controlled activities become possible. First, the effectiveness of instructional/prompting activities can be enhanced by enabling the computer to provide reinforcement following successful task completion. Reinforcement can take the form of video displays, such as are utilized in video games and programmed instruction, verbal praise and encouragement via synthesized speech, or even tokens through the utilization of a computer controlled dispenser. Additionaly, systems can be configured to provide corrective instructions when errors occur, pace work speed according to error rate, determine slowdown and other deterioration in performance, and schedule rest periods or summon a human supervisor to evaluate the problem. Conceptually this model is as straightforward as computer based arcade type games which monitor task performance consisting of game paddle or joystick manipulation. Certainly, there are technical problems in attempting to monitor task performance on more complex work activities. However, this is far from an impossible situation. The same technology which is used in automobile engine monitoring systems and enables robotic systems to detect and respond to environmental change can be utilized in monitoring work activities involving assembly, disassembly, sorting, etc.

The workings of a model system are depicted in the following example which is designed around a parts sorting task. The specific job is to sort bolts and nuts from quantities of other small parts and to put them in separate containers. The system components consist of an Apple IIe class computer, color monitor, interactive videotape player system, speech synthesizer, headset through which the aural messages are given to the worker, a supervisor alert device, e.g., a flashing light over the work station, a token dispenser, and a work task monitoring system. The latter is comprised of an electronic digital scale on which the miscellaneous parts container rests, and a bank of ports into which the sorted parts are to be placed. The three ports, which are for the bolts, nuts, and other parts, have computer activated covers, and internal switches which are tripped when parts fall through them. All are interfaced with the computer.

When the system is activated, a synthesized speech greeting is issued to the worker through the headset and the instruction/prompting sequence begins in the form of aural and video displayed messages, for example, "take a part from the container." At that moment, a timing sequence is started by the computer. When a part is removed from the container, the change in weight detected by the scale is input into the computer, which stops the timing sequence and files the amount of time which elapsed. If a part is not removed by the time a preset parameter has been reached, a reprompt is issued. If several successive reprompts are issued and no response by the worker has occurred, the supervisor alert system is activated. After a part has been removed from the container, the computer then determines the kind of part selected by calculating its weight, that is, subtracting the new container weight from the old and comparing it with a programmed weight list. Once the nature of the part is determined, a second sequence is started. Aural and visual instruction is given to the worker to place the part in the appropriate port, e.g., "now put the bolt in the blue container." The appropriate container lid is activated/opened on computer issued command and interval timing begins. Again, reprompting may occur if necessary, and when the part falls through the port, the timing is stopped, and a praise message, "good," is issued to the worker. The whole cycle then repeats. Other program elements enable the prompting sequence to be speeded up to an optimal level, and to alert the supervisor when performance begins to deteriorate. Additionally, the system can be configured to alert the supervisor when errors occur, for example, if two parts are removed from the container without the first being placed in the appropriate port. Also, when a predetermined number of successful sorts is attained, the computer causes the dispenser to emit a token and a video display of accumulating tokens is presented together with an enhanced praise routine. Upon termination of the activities, the accumulated performance information is analyzed by the computer and a report is generated.

IMPLICATIONS

What has been presented is a currently feasible model of computer aided supervision which can be configured to meet the needs of persons with varied types of cognitive impairment. Ulitization of such a system can help to enhance the productivity of workers with these problems. In the optimal situation, it may be possible to enhance performance to the extent that these workers can function and compete in the regular work place. Should that not be attainable, utilization of the system may enable more severely handicapped persons to function in sheltered settings or it may enhance the efficiency of the sheltered workshop, resulting in the need for less subsidization. The factor which makes this system a viable alternative to human intensive supervision is that the core is the typical personal computer available at increasingly nominal cost. The primary challenge is development of task monitoring systems.

Perhaps the greater issue at this time is the fact that such systems have obvious potential for abuse. Basically, the worst possible scenerio presents a picture of dehumanization and exploitation. An extensively controlled work environment immediately precipitates "Orwellian" pictures of mind control and workers reduced to zombie-like parts of a production line. This is not a farfetched notion in that such a system can be configured to provide aversive conditioning, as well as positive reinforcement to shape behavior.

Moveover, if utilized under the most benign circumstances, there is need for concern. Specifically, at this point in time, the psychological effects of functioning in such a controlled environment are unknown. What might be the effects of interacting so intensively with a controlling machine for eight hour days, five out of seven days a week? How might the social skills of this population be affected the situation? Obviously, these and other related issues must be satisfactorily addressed before aggressive advocacy of this model is initiated. Already, with this sort of close monitoring, e.g., in the information processing area, philosophical, "big brother" issues are being raised by workers. On the other hand, successful utilization of the system holds the potential to enable a sizeable handicapped population to attain greater economic self-sufficiency, as well as enhanced dignity and self-esteem which derive from being a productive member of the community. On a macro level, the system has potential to reduce the financial burden borne by society to maintain less productive citizens at an acceptable level. The potential benefits of successful development and implementation of computer aided supervison to enhance the productivity of persons with severe learning handicapps warrant serious pursuit, but careful consideration must be given to these moral/ ethical and other issues to insure that the technology does not harm those individuals whom it is intended to help.

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2. Lynch, K.P. Toward a skill-oriented program for trainable and severely mentally impaired adults. In Bellamy, G.T., O'Connor, G., & Karan, O.C. (Eds.). Vocational rehabilitation of severely handicapped adults. (pp. 85-101). Baltimore: University Park Press, 1979.

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1.2

DEVELOPMENT OF A MOBILE FACILITY TO PROVIDE ACCESS FOR RURAL SUBJECTS TO AN EVALUATION SYSTEM

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ABSTRACT

Apparatus for the administration of the Available Motions Inventory, an evaluation of physical capacity, was installed in a van. The feasibility of achieving an installation which would allow for the comfort of the evaluation subject and the evaluator as well as to provide adequate protection for the apparatus in transit was investigated. A satisfactory ststem was designed and fabricated. When the installation was complete, a primary purpose was carried out, that of comparing the circumstance in which the evaluation apparatus was transported to the subject's home or agency to the circumstance in which the subject would be transported to the urban center where the evaluation was typically administered. The mobile format was determined to be economically feasible only when more than one subject was evaluated at one location. Pragmatic feasibility was established for young or severely disabled adult subjects for whom transportation was not readily available.

INTRODUCTION

The Available Motions Inventory was developed by the research staff at the College of Engineering, Wichita State University, in 1972. Extensive modification and refinement have been carried out by the Rehabilitation Engineering Center at Wichita under the joint sponsorship of the Cerebral Palsy Research Foundation of Kansas, inc. and the College of Engineering at Wichita State University. The primary purpose of the Available Motions Inventory, AMI, is to document the physical capacities possessed by a physically impaired person, hence the term "Inventory". The analysis of the AMI originally focused upon the application of the identified abilities to perform work in an industrial setting. Refinement of the analysis system allows for the abstraction of the repertoire of physical mini-tasks so that it can be applied to alternate settings, such as white collar occupations, activities of daily living and educational environments.

The Rehabilitation Engineering Center at Wichita has utilized the data realized from the Available Motions Inventory to bracket given individuals' occupational potential as well as to analyze the alterations which would be required to render a particular worksite operable by a particular disabled worker. Information derived from the administration of the AMI has been utilized to place disabled persons in sheltered, semi-sheltered and mainstream settings.

BACKGROUND

Need

More than three hundred disabled persons have been evaluated at the Wichita site. Many have traveled/been brought to Wichita at great cost. Kansas is a primarily rural state, having three urban centers, all located in the eastern half of the state. It was deemed useful, therefore, to investigate the feasibility of rendering the AMI evaluation accessible to persons living in areas remote to Wichita. This was to be achieved by installing the apparatus in a van in such a manner that the evaluation could be administered in the vehicle.

Target Population

Three general classes of subjects were considered as having a high potential of benefitting from the mobile evaluation format:

1. Subjects requiring the support of medical personnel or equipment.
2. Several subjects residing and/or participating in a day program at a single location.
3. Subjects whose potential for altered programming was marginal.
4. Subjects for whom transportation to the urban evaluation site was not readily available.

MOBILE EQUIPMENT

Vehicle

The van which was acquired for this project has a large cargo box, 7 feet by 7 feet by 14 feet long, Figure 1. This space allows for a room-like environment in which to administer the evaluation. A heavy duty lift was installed which provides access for wheelchair bound individuals, Figure 2. Insulation was installed between the outer shell and the inner wooden liner. An air conditioner/heater unit was mounted on the roof. It has been proven possible to utilize the vehicle when temperatures are as low as 35° outside.

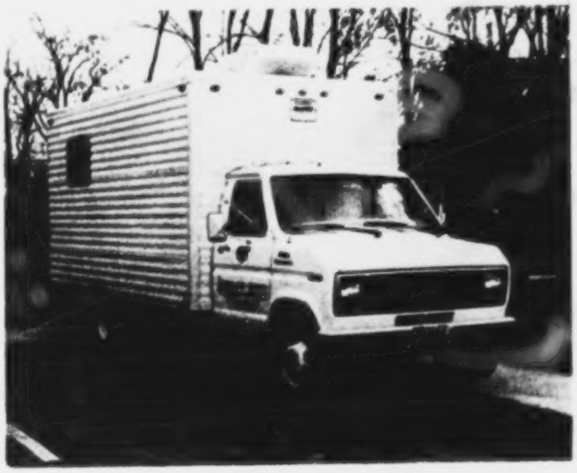

Figure 1

Mobile AMI Van

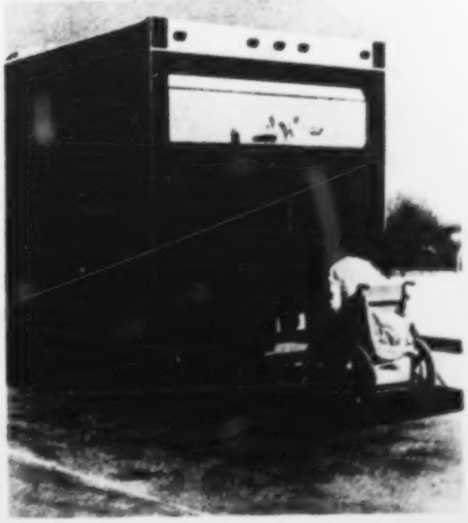

Figure 2

Lift Acess to Mobile Van

Florescent lighting was installed. Power for the vehicle is provided by two 110 volt circuits. They are connected by heavy power cords at the evaluation site. A non-intrusive decor was created to further enhance the room-like atmosphere. The carpeting is heavy duty, low napped, to allow a wheelchair to roll readily on it.

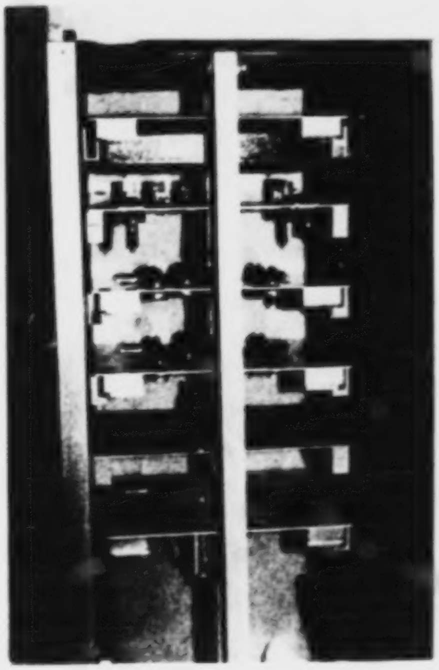

Figure 3

Equipment Storage

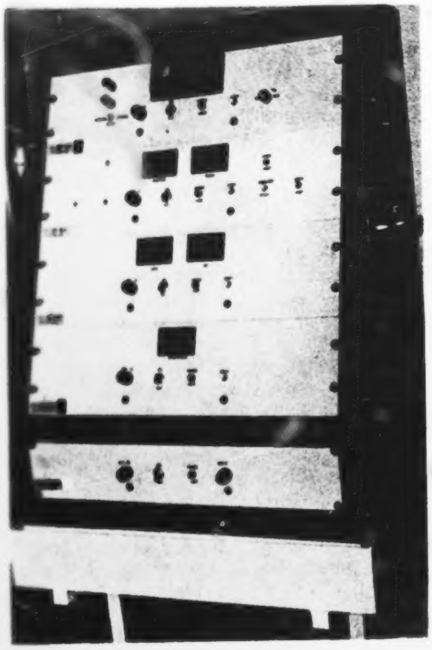

Figure 4

Instrument Console

Evaluation Apparatus
The frame for the evaluation modules was attached to the wall of the van. A motor-driven mechanism was provided to adjust the height to the specifications of the individual subjects. Storage cabinets were custom designed to secure the equipment in transit, Figure 3. A single console was designed to house the instrumentation required for the evaluation, Figure 4.

METHOD

Equipment Evaluation
On site check-out of the system was carried out by evaluating able bodied subjects. An initial trip was undertaken to verify the stability of the test equipment. Three disabled subjects, two of whom were wheelchair bound, were evaluated. The van equipment and the evaluation apparatus functioned as required for successful administration of the evaluation procedure.

Subject Identification
Nomination of potential subjects were solicited from internal and external agency contacts of the Cerebral Palsy Research Foundation of Kansas, inc. Twenty-five subjects in all were evaluated utilizing the mobile format in a remote location. Nine were contacted through other projects of the Rehabilitation Engineering Center. They were subjects from public school classrooms and a cooperative activity with a Projects With Industry, PWI, agency. Eleven subjects were proposed by the SPARK project, Special Population Activities for Rural Kansans. This agency serves a large rural area in which physically disabled persons are very sparsely distributed. Specialized programs are not

readily available in such areas. SPARK was instrumental in making contacts with three sheltered workshops and a large consolidated special education cooperative. All four of these agencies' programs are ones in which clients or students are gathered in to a program at a central location.

RESULTS

Of the twenty-five subjects evaluated, only one was evaluated singly. Ideal numbers of subjects at a single location were two, who could be evaluated in a single day; and four or more, optimizing the per diem spent for trips of more than one day. Six additional observations were performed on subjects deemed too severely impaired to perform the AMI. The purpose of such observations was to bracket the potential performance possibilities.

DISCUSSION

Small economic advantage was realized from utilization of the mobile format with this group of subjects. The primary advantage realized from this investigation is to render the AMI evaluation accessible to persons who could benefit from its use. Over half of the subjects (14/25) did not appear to have the potential to travel to the urban location where the evaluation is usually administered. Deterring factors were personal economics and low expectation of altered programming as a result of performing the evaluation. Over one third (9/25) of the subjects were identified as having the potential of enhanced occupation if work site adaptations were carried out. Should the agencies decide to have such alterations performed, the AMI data would provide the basis for such modifications. An additional benefit realized from the interaction with subjects and professionals in these remote areas was to establish liasons with resources for highly specialized theaapies not readily available in rural areas.

CONCLUSIONS

There appears to be merit in rendering rehabilitation evaluations and therapies mobile in order to serve disabled persons in rural areas in a more equitable manner. This appears to be consistent with the principle of equal access to strategies and services designed to alleviate the penalties imposed by disability.

ACKNOWLEDGMENTS

The authors wish to express their appreciation to Ms. Karen Baron, director of SPARK, Manhattan, Kansas, for her assistance in identifying rural subjects for this project.

The authors wish to acknowledge the exceptional design and fabrication work performed on the AMI vehicle by Ron Ingalls, Design Engineer and Steve Mallernee, Technician.

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USE OF A SMALL ROBOTIC ARM IN THE WORKPLACE BY A SEVERELY PHYSICALLY DISABLED INDIVIDUAL

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ABSTRACT

The Rehabilitation Engineering Center in Wichita, Kansas, has, as its emphasis, research in the area of finding means to place severely physically disabled persons in the workplace through the use of technology. A project to investigate the use of small robotic arms to assist the disabled at work has been underway for two years at the time of this writing. One work station at Center Industries Corporation has been identified as having good potential for the use of such a robotic device and is being manned by a young man who has cerebral palsy. This work station and the results of the use of the robotic arm are presented.

INTRODUCTION

The Rehabilitation Engineering Center (R.E.C.) in Wichita is federally funded by the National Institute of Handicapped Research and is also under the sponsorship of the Cerebral Palsy Research Foundation of Kansas, Inc. One of the 13 research projects that the R.E.C. is currently investigating is evaluating the use of small robotic arms in aiding the severely physically disabled in the workplace. Center Industries Corporation in Wichita is a company which employs disabled persons in the "real world" environment of non-sheltered employment. A contract which Center Industries is currently fulfilling is that of doing pre-solder tinning of electrical component leads for the Boeing Company. It is on this line that a work station was chosen to first try such a robotic device.

METHODS

The robotic device in question is of the type that maintains a base position but has all of the motions that the human arm is capable of performing. Figure I portrays an artist's conception of the degrees of motion of a robotic arm. To the "hand" of the device shown in Figure 1 must be added an "end effector" or "gripper" to facilitate the handling of parts or materials.

The particular robotic arm that was chosen and purchased for this project is the "Teachmover" which is offered by Microbot, Inc.^{1} The robot, along with the work station can be seen in Figure 2. This particular device was chosen because of its high degree of accuracy and repeatability and low cost ($2,600).

The tasks of the job chosen are: 1. Picking up the part; 2. Immersing the wire lead of the part in a liquid flux; 3. Immersing the wire lead of the part in molten solder to within 1/32 inch of the body of the part; 4. Reversing ends of the part and repeating the process. This operation is usually done by hand as shown in Figures 3 and 4.

The robot is programmed to progress through the required tasks of the job by manually operating the "arm" and "hand" using a teaching pendant as shown in Figure 5. The microprocessor which is contained in the base of the robot "remembers" the various steps in the cycle and then repeats the cycle when instructed by a single key activation on the teach pendant. The disabled worker in this case is able to start and stop the robot cycle by pushing the single keys required to do so.

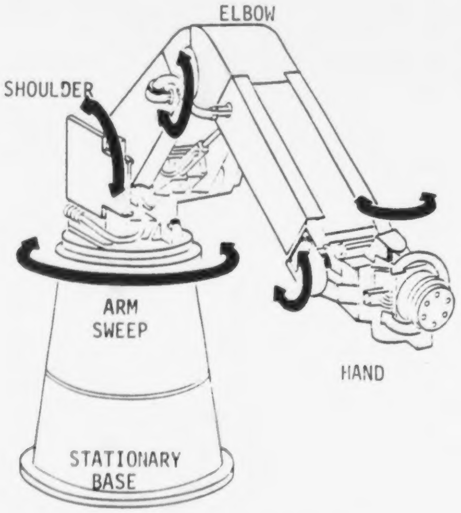


FIGURE 1: ROBOTIC ARM

FIGURE 2: WORK STATION

The worker is needed at the work station to perform the task of indexing parts between cycles so that the robot has the same position from which to pick the next part, and, to visually inspect the solder coating on the wire leads to insure quality.

Figure 2 shows the worker sliding a single part on a slightly elevated platform to a slot in the platform. Below the slot is a slide chute at the bottom of which is the holding fixture or indexing device from which the robot picks the part. Figure 6 shows the holding fixture at the end of the slide chute.

The robot, after picking the part, then moves to immerse the wire lead of the part in the flux (Figure 7) and then to the molten solder which is contained in an electrically heated pot (Figure 8). After immersion in the solder of the first wire lead the robot then places the part in a second holding fixture for the purpose of turning the part end-for-end. This is necessary because the depth of the immersion is held to a close tolerance and the robot indexes the depth by locating from one end of the body of the part at a time. See Figure 9. The robot deposits the part in a solution of alcohol and the lead person in the shop recovers the parts periodically to pass on to shipping.

DISCUSSION

It has been determined that use of such a robotic arm has enabled this worker to produce at a rate similar to his able-bodied peers with equal, or better, quality. Some problems have been encountered, however. The particular robot chosen operates with "stepping motor" technology. If the arm of the robot is "bumped" the drive motors will "skip" some steps thus causing the cycle to be disrupted and the robot must be reprogrammed before going on. Also, this particular robot has had periods of erratic movement thought to be caused by the microprocessor somehow "sensing" some electrical transients from the electrically heated solder pot as its power is controlled on and off by a thermostat. Electrical shielding is being investigated and it is anticipated that this problem will be solved shortly.

CONCLUSIONS

Observation of this particular worker using the robotic arm to perform the tasks of this particular work station has assured the investigator that small robotic arms indeed have a significant place in the vocational placement of severely physically disabled workers.

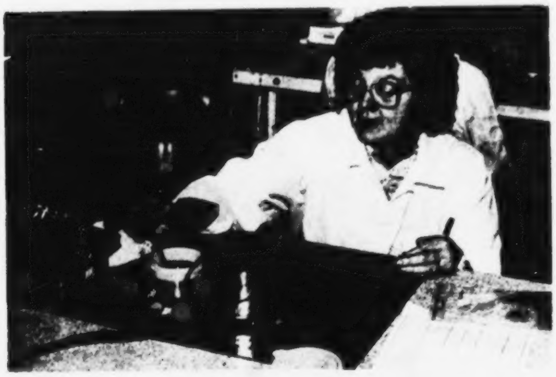


FIGURE 3: IMMERSION IN FLUX BY HAND

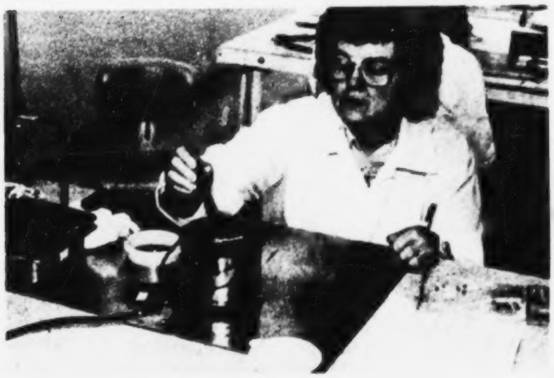


FIGURE 4: IMMERSION IN SOLDER BY HAND

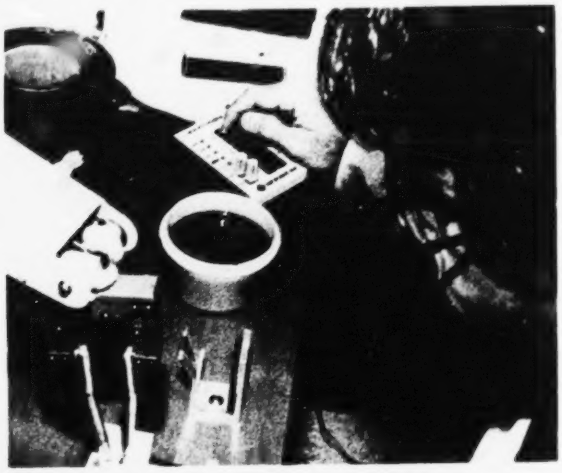


FIGURE 5: PROGRAMMING WITH TEACH PENDANT

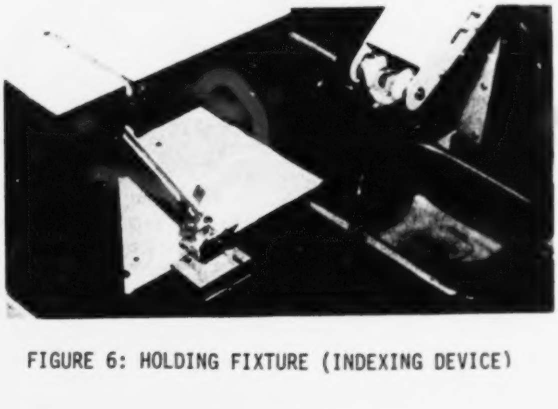


FIGURE 6: HOLDING FIXTURE (INDEXING DEVICE)

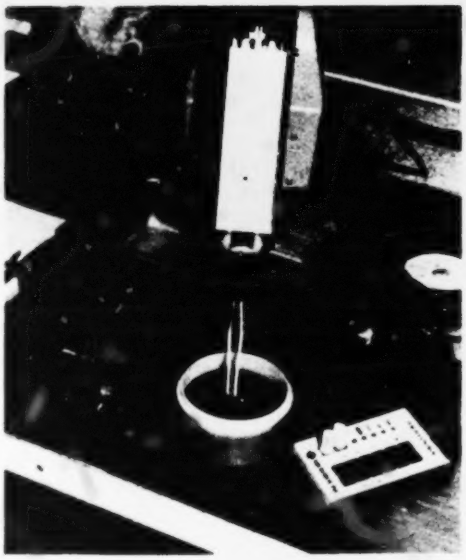
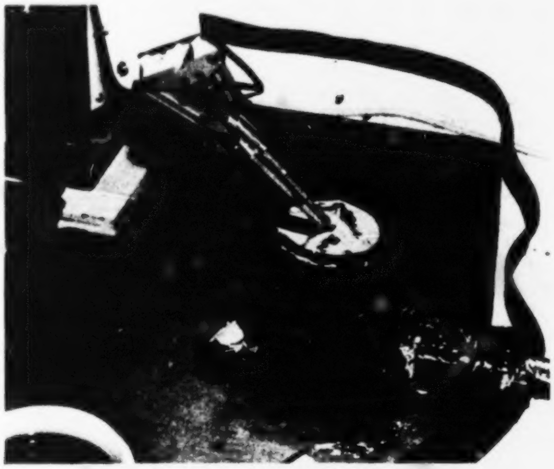


FIGURE 7: IMMERSION IN FLUX BY ROBOT



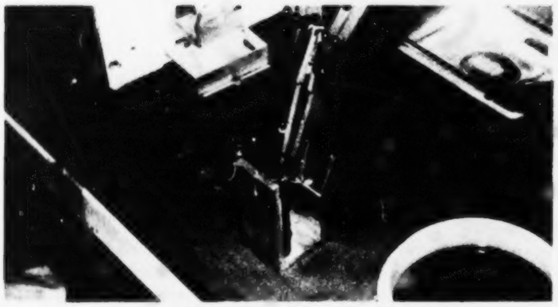


FIGURE 9:
HOLDING FIXTURE FOR TURNING PART END-FOR-END

ACKNOWLEDGEMENTS

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FOOTNOTES

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1.4

ROBOTS IN LONGTERM HEALTH CARE

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ABSTRACT

An interdisciplinary team was selected to participate in an investigation of robots in a longterm care facility. The research took place in the VA Menlo Park Nursing Home Care Unit. The objectives were three-fold. First, we wanted to identify tasks that were amenable to augmentation or replacement by robotic technology. Second, we wanted to survey the attitudes of the administration and staff toward computers and robots. Third, we wanted to see if our training techniques for the VA/SU Robotic Aid could be used successfully by older users and to see how their time to completion scores compared to younger users.

BACKGROUND

A recent business journal article stated, *For a health care industry looking to cut costs while retaining high margins, robots represent a new and practical technology that allows the automation of processes that simply could not be automated previously and that could provide a competitive edge for forward-looking companies.* [10]

Health care expenditures in this nation account for 10.5% of the Gross National Product. They were approximately $350 billion in 1983. [5] These costs have been rising relentlessly each year -up from 4.5% in 1950. It is instructive to note that, at present, approximately 50% (39-60% depending on hospital type) of most hospital's expenditures are labor related [1] with about one-half of that sum going to nursing costs. Robots as aids to nursing staff have cost-cutting potential. Leifer states that *robots, at an estimated $6.00 per hour, could reduce the cost of attendant care which now has a net cost of as much as $15.00 per hour.*[9]From the public's perspective, this is a relevant policy issue. This year about $40 billion dollars will be paid to hospitals for medicare–about 38% of hospitals' overall income.[3]

Two **trends**, (a)the push toward more in-home care for relatively sicker patients, and (b)the entry into longterm care facilities by older, increasingly more frail persons, offer unique opportunities for the development of new approaches for using robotic technology. The workload on the caregivers, in homes and in institutional settings, will become increasingly demanding. There will be more need for devices that can serve, lift, transfer, ambulate, remind, fetch, carry, monitor, amuse, and communicate. Considerable savings can be effected by integrating as many functions as possible into one operational unit— in short, a robot.

People will be discharged from the hospital sooner and, therefore, may require more care at home[4]; The advent of these sicker patients at home has produced a growing home health care market. *Home health services for private patients collected $4.2 billion in 1981, the last year for which figures are available, according to market researchers Frost & Sullivan Inc. Insurers paid $2.3 billion of this sum.*[2] This market is also being spurred on by those insurance carriers who are demonstrating the cost-effectiveness of home health care. Aetna Life & Casuality calculated a net savings of $40,761 per month if a baby born with breathing and feeding problems is treated at home rather than in an acute care hosptial. Likewise, home health care also yielded a savings of $9,931 per month for a quadriplegic spinal cord injured patient.[2] Home- and community-based care is the fastest growing component of our health care system. *Prospects are equally bright for manufacturers and distributors of health-care products geared toward ambulatory or home-bound patients.*[12]

The second major trend is that people are living longer with disabilities and therefore, require more health care in general. Persons over the age of 85 constitute the fastest growing subgroup of our U.S. population. Skilled Nursing Facilities total revenues, for 1981, topped $24 billion. This growth (in the over age 85 citizens) will increase the need for institutionalization, but the patient profile will change as age-debilitated individuals, requiring more intense levels of care, become the majority of the residents of longterm care institutions. **Longterm care** is one of the most 'troubled and troublesome' segments of the health care market. This is a fertile area for creative problem solving. Both in-home and institutional settings are open to innovative solutions to reduce costs and improve quality of care. These two trends in health service delivery can help us begin to define specific areas of need in which robotic technology can be used to address a wide range of potential applications. U. S. health care delivery may, for our purposes, be viewed as a continuum of health care service areas [8] in which robot technology might be utilized to varying extents.

PHASE 1: TASK IDENTIFICATION

METHODOLOGY

An applications research team was formed to find areas in which robot technology would fulfill a need. Observations were carried out in the Veterans Administration Nursing Home Care Unit (NHCU) in Menlo Park, California. The team was composed of an interdisciplinary group of fifteen individuals with 25 years collective experience using the VA/SU robot. The team was composed of the following members:

K.G. Engelhardt, AB, BA, SNF/Robot Study Director, Evaluation Supervisor, RR&D; **Roger Awad-Edwards**, MS, SNF/Robot Study Coordinator, Health Services Researcher at RR&D; **Paul Haber**, MD, Regional Coordinator on Aging, Veteran Administration; **Machiel Van der Loos**, Ing., SNF/Robot Study Head Engineer, Mechanical Engineer at RR&D; **Dwight Wilson**, MS, Nurse Supervisor, VA Nursing Home Care Unit, Menlo Park; **Margo Apostolos**, Ph.D candidate, Stanford University, Education; **Gayle Curtis**, MS, Systems Engineer at RR&D; **Merry Lee Eilers**, MA candidate, Stanford University, Medical Anthropology; **Vicki Ellis**, MS, Nurse Gerontologist,VA Nursing Home Care Unit, Menlo Park; **Karen Holloway**, BS, Design Engineer, RR&D; **Brodie Lockard**, MA, Interactive Educational Technology Consultant, RR&D; **Nick Pruneda**, SCI veteran, RR&D; **Walter Vogel**, MD, retired internist, volunteer, RR&D; **Wayne Young**, retired production engineer, volunteer, RR&D; **Eleanor Zielski**, PHN, VA Nursing Home Care Unit, Menlo Park;

During phase 1 researchers were paired with individuals of dissimilar backgrounds (for instance, an engineer with an anthropologist) to *silently observe* activity in the NHCU during various shifts over a one week period. Pairs were rotated and time of day for observation was varied. Researchers were instructed not to interact with staff but to become familiar with the skilled nursing facility environment and to look for areas of need that robotic technology might address.

In the second round of observatons, during the second week of the study, researchers were permitted to discuss their observations with staff. NHCU Staff also contributed their ideas for applications of robot technology during the two staff orientations.

RESULTS

In the area of robot applications, our team was able to identify twelve major categories with fifty-four sub-groups. The major categories for these tasks are shown below.

Potential Robot Tasks:

transfer-lift-transport
housekeeping
ambulation
physical therapy
depuddler (urine cleaner)
surveillance
physician assistant
nurse assistant
patient assistant
mental stimulation

PHASE 2: ATTITUDINAL ASSESSMENT

METHODOLOGY

The second stage of our research examined, by questionnaire, attitudes of staff and administration in the Veterans Administration NHCU.

The questionnaire for the attitudinal assessment includes questions from studies conducted by our research team as well as by Sanford Dornbusch, Ph.D., Stanford University and by Iseli Krause, Ph.D., Syracuse University. It contained 111 questions of different types: Likert-style, open-ended, and dichotomous (yes/no), and semantic differential. It was administered to the NHCU staff as part of our standardized orientation.

RESULTS

Of our 22 respondents, 3 were male and 19 were female. Their average age was 45 years. Sixteen had nursing related staff titles and the remaining six had other health and human service professional staff titles. Highlights of their responses include:

- None use a computer at work.
- 95% agreed with the statement, *Computers have made living easier.*
- 86% agreed with the statement, *A robot could be useful in my job.*
- 96% agreed with the statement, *Robots have many beneficial applications.*
- 100% agreed with the statement, *Robots can be used to serve people.*

The results of our research indicate a very positive response to computer/robot technology. In general, all respondents described robots as ***positive, valuable,*** and ***useful***. The Clinical group were very enthusiastic toward human service applications of robotic technology. Informal surveying and interviews with the patients who participated indicated unexpectedly positive responses.

PHASE 3: FEASIBILITY

METHODOLOGY

The methodology for demonstrating the feasibility of robots in a nursing home followed the Interactive Evaluation Model already implemented at RR&D. [6][7] Standardized training procedures were utilized to ensure that NHCU patients and staff could demonstrate comparable proficiency in the use of the VA/SU robotics aid. Our training approach employs an adult-learning theoretical framework, known as ***andragogy***. Andragogy is based on the concepts of self-directedness, learning readiness, immediate applicability, and problem-centered learning tasks.

RESULTS

Two in-patient residents and two staff members (a nurse supervisor and a social worker) were trained on the VA/SU robotic aid. The two patients were male, ages 62, and 90; the nurse supervisor was male, age 35; and the social worker was female age 39. Each successfully completed at least two sessions with the robot. All users were able to complete our first standardized training task (picking up a cup–with straw– and giving themselves a drink) within one standard deviation of our mean task completion time of two minutes. (Subsequent to our study, two more patients were trained: male, age 61 and female, age 87.)

For the first time in history, the study established the feasibility of successfully training older persons, in a longterm care facility, to use a robotic aid.

FUTURE

Potential robot task areas exist in nearly all health care institutions. We can now begin to prioritize the task categories and develop methods for examining robot diffusion at any point along the health care continuum and identify (1)human needs, (2)tasks and task sets, and (3)task and robot characteristics for each environment or setting. This potential robot task list can be generalized to a wide range of health and human service settings.

Potential Health Care Settings:

emergency
hosptial
skilled nursing facility
intermediate care facility
domiciliary
residential
in-home
ambulatory (outpatient)

Our work will continue to examine robotic technology as it evolves and the role it can play in human independence. We anticipate a continuing move toward the merging of industrial and personal robot technology. We expect voice control and sensor technology advances to play pivotal roles in future widespread applications of robotic technology.

We have demonstrated the feasibility of using robotic technology to serve severely disabled and older people. The Veterans Administration is continuing its vanguard role in researching the utility and potential marketability of robotic assistants as the technology's capabilities expand.

CONCLUSION

The blending of medical expertise and engineering know-how, combined with interactive evaluation throughout the product's life-cycle and industrial interests, could result in assistive robots that have the potential for improving the quality of life and increasing the independence of long term care patients. The time has come to bring our older citizens into our technological society of the 1980's. The challenges to the sociotechnological system are enormous. Innovative designs which transcend age and ability constraints, and applications of state-of-the-art technology to humans of widely diverse ages holds the promise of helping us create a forgiving environment in which all humans can thrive.

ACKNOWLEDGEMENTS

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DEVICE TO FURTHER INDEPENDENT LIVING

Anne W. Leung
Winfield R. Matsler

ABSTRACT

The Independence Lift is a device which allows either a weak paraplegia, a quadriplegia with little motor skills, or a person otherwise requiring an attendent to lift himself or herself out of a chair and into a bed, or out of a bed and into a chair.

Construction materials consisted of two aluminum poles with feet and wooden bases as stands, a steel black pipe that spans between both poles, and a section of aluminum I-beam which was attached to a steel black pipe. For horizontal movement, a trolley was a fixed to the I-beam and is operated manually. Connected to the trolley is a electrical motor for vertical movement. Under the hoist are two straps, one for the shoulders and the other for the legs.

There have been both psychological and health benefits as a result of this system. Furthermore, it can be expanded for more uses in the home and institutions.

INTRODUCTION

The Independence Lift was developed for a transitory student who is a quadriplegia with some motor skills in his arms. In the past he has required an assistant to lift him in and out of bed onto his wheelchair.

Although there are a limited number of lifts on the market for use in the private sector, none have been found adequate for his use. Furthermore, these lifts may have been employed by the client, but they are not adaptable to his present housing situation, and the harnesses were totally inadequate.

This paper will further discuss the client's background, the materials and methods used in construction, results of the product, and a discussion on further expansion of the system.

BACKGROUND

This project was undertaken for a fellow student at Southern Illinois University who was in a diving accident. As a result of this accident he was rendered a quadriplegia, with full strength of the neck, 5 lbs. lifting power in the right bicep and 10 lbs. in the left bicep.

He has made the decision to live alone in his strive for independence. An attendent is, however required twice a day to lift him in and out of bed. This has been both a psychological and financial strain. Having the attendent for such a limited time poses the problems of relieving pressure sores, getting out of bed in emergency situations and taking naps at will.

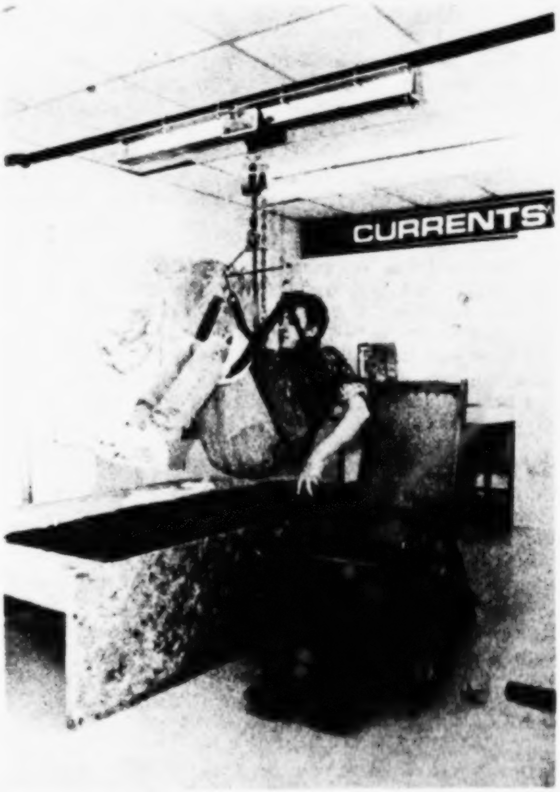
Although there are lifts for this purpose on the market, identification of one the suits his needs has not been made.

MATERIALS AND METHODS

After the initial design was completed the purchase of the necessary materials was required with limited personal funds. With the basic design in mind materials were sought from various salvage yards.

Two 2" I.D. aluminum volly-ball posts, with three prong feet, a 4'x4"x4" aluminum I-beam and a 1¼"x 9' steel black pipe were found that were adequate for the super structure.

The I-beam was U-bolted to the steel black pipe, and attached to the posts forming the basis frame. In order to stabilise the structure, the feet were bolted to wooden bases. A trolley was suspended from the I-beam which would enable the client to move horizontally. A manual hoist was attached to the trolley, as opposed to an electric hoist because it is considerably less expensive and would enable the client to build strength in his biceps.



Early developmental stage

Winfield R. Matsler designer of the Indepencence Lift. Early developmental stage photo taken in the Design Department at Southern Illinois University at Carbondale.

Existing sheets from bathing hoists were utilized but proved inadequate for the client has not the ability to balance himself, and would actually roll within the sheets. This harness was also cumbersome for he had to sit on them constantly, and also when in use they restricted his arm movement rendering him incapable of operating the lift. Therefore, the next task was to design a harness that would solve these problems. The solution consisted of using two straps padded with egg crate foam employed as the harness. One to support his legs the other his shoulders. The shoulder strap is a loop that he can slip over his head and under his arms. One end of the leg strap is connected to the hoist, and needs to be manuvered under his legs and connected to the hook on the hoist.

The harness had one drawback in this application in that when suspended for more than twenty minutes, which was the amount of time required to manipulate the manual hoist, the client's blood circulation was restricted.

In this case there was no alternative but to purchase an electric motor, that would reduce the time of suspension. A rechargeable 12 volt D.C. motor which only requires two minutes, and would remain in service during power failures was chosen.

Once suspended he slides horizontally by manually pushing off of the wheelchair and grabbing onto the sheets of the bed.

RESULTS/DISCUSSION

The Independence Lift is not a generic solution, but it is adaptable. If a client has the ability to use a manual hoist it should be employed as to help build upper-body strength and would also be more cost efficient. The trolley, if necessary, could also be motorized.

Psychologically he experiences a higher degree of confidence and self-sufficiency. As a result of this heightened sense of self-esteem, the client has found the initiative to pursue other goals toward total independence.

The client is now capable of transporting himself in and out of bed, enabling him to relieve pressure sores, nap at will, and gain a higher level of mobility in the event of emergency situations.

Federal and state agencies and other non-profit organizations could possibly implement the system, resulting in lower costs of subsidizing health care. Furthermore, it may have various uses in hospitals and rehabilitation facilities.

CONCLUSION

The Independence Lift has proven sucessful for the client which it was specifically designed for. It has helped in furthering his independence and also has psychological and health benefits.

The end product was the result of assessing inadequacies in existing lifts, and a practical and economical design.

The structure's simple, resourceful design will result in long life use with basic maintenence.

ACKNOWLEDGMENTS

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AUTOMOBILE DESIGN FOR PHYSICALLY DISABLED AND ABLE-BODIED POPULATIONS

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ABSTRACT

Archival, interview and experimental research have concluded that an automobile system suited for the independent driving of both able-bodied and physically disabled populations will be acceptable by both. An automobile design that will satisfy the independent driving needs of both able-bodied and physically disabled populations can be mass produced in quantity to satisfy high consumer demand. The result will be an automobile design that is more affordable to the physically disabled than existing "disabled persons" automobiles.

This paper and accompanying graphic material describe the research based design of an affordable mass producable automobile which can be entered and operated by both physically disabled and able-bodied populations.

RESEARCH

Archival, interview and experimental research were used to develop and vertify an acceptable concept for the design of a transportation device which will satisfy the independent driving needs of special and able-bodied populations.

Developing a Design Concept. Archival and interview research representing phase one of the research process was used to formulate a concept for the design of a vehicle for the independent transportation of the physically disabled. The following topics were explored in this phase of the research: special populations and their physical, psychological, demographic, and financial characteristics; existing vehicles and adaptations intended for use by special populations; vehicle safety; federal vehicle regulations; characteristics of able-bodied populations; and human factors data for both physically disabled and able-bodied populations.

These findings concluded that most attempts to provide independent transportation for the physically disabled have resulted in costly vehicle alternatives that satisfy only a small percentage of all those that are handicapped.

The physically disabled presently have two options for independent transportation. One alternative is to purchase a vehicle that has been specifically designed and manufactured to accomodate a narrow segment of the physically disabled population. Because these vehicles are intended for only a small percentage of the physically handicapped, their production runs are quite low in number. As a result, these automobiles are extremely expensive, thus reducing their availability to a very narrow segment of an already small group of people.

The second alternative for the physically disabled is to "customize" a factory available vehicle (designed for able-bodied populations) by installing adaptive controls and modifications to suit almost any physical limitation. Although these modified vehicles are better able to meet the needs of a diverse range of physically disabled populations, such installations are expensive and the final solution is always based on a design intended for able-bodied populations.

Both alternatives are worthwhile attempts to reduce the mobility limitations of handicapped populations. Restrictive costs and unaccomodating vehicular design have prohibited many physically disabled from experiencing the freedom associated with independent transportation.

These conclusions led to a logical approach for an automobile that will satisfy the independent driving needs of physically disabled and able-bodied populations. By devloping one vehicle design which can be entered, exited, and controlled by both populations, production quantity will increase, thus decreasing consumer cost. The final result will be an affordable, mass producable automobile which will accomodate almost all populations.

Verification of the Design Concept. Phase two of this research was to determine the validity of this concept. The question, "Will able bodied populations purchase an automobile design which would also be owned and operated by physically disabled populations?" was answered with experimental research.

Three surveys were used to manipulate the significant variable of this experiment (able bodied preference for vehicles that are and are not associated with physically disabled populations). Survey one asked recipients which vehicle they preferred, Vehicle A or B without indication that either was intended for use by the handicapped. This survey was used to establish a control group to act as a basis for comparison.

Survey two identified Vehicle A (Vehicle A and B are the same as those in Survey one) as the "handicapped" automobile. Able-bodied recipients were then asked which of the two vehicles they preferred.

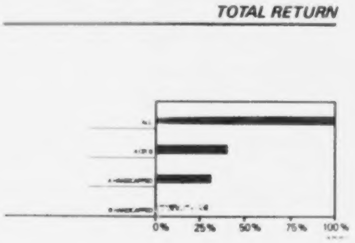
Survey three identified Vehicle B as the "handicapped" vehicle, and then asked which of the two vehicles were preferred (both Vehicles A and B were the same vehicles as in surveys one and two).

The results of Survey one were to be compared with the results of Surveys two and three to determine if a vehicle that was identified as adaptable and available to handicapped populations would be acceptable for ownership by the non-handicapped public

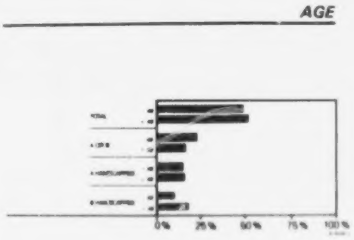
BEST COPY AVAILABLE

Each survey was sent with an accompanying cover letter describing the purpose of the study, and an addressed and stamped return envelope. Each of the three surveys was sent to a randomly selected group of fifty people (one person in each state) to obtain 150 total recipients. One geographic location was randomly selected in each state. Surveys one, two and three were sent to three separate people within that location to eliminate some of the variables that may have affected the study results.

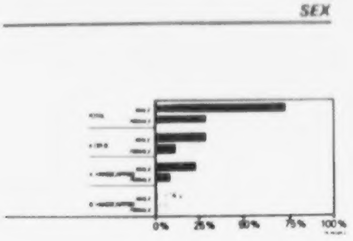
The results of this research conclude that able bodied populations are more likely to purchase an automobile intended for both physically disabled and able-bodied populations if able-bodied populations are aware of this flexible design feature, if that automobile is comparable in price with other similar vehicles and if the automobile meets user needs and specifications. (See graphs 1-4 for a more detailed explanation of these results.)



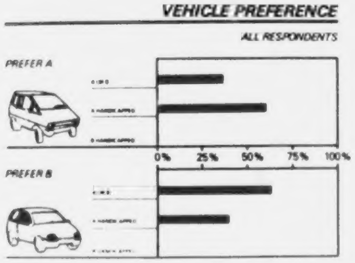
Graph 1. Total Return - of the 150 surveys sent, 75 were returned. Of the 75 returned, 28% returned Survey one (A or B); 31% returned Survey two (identifying A as "handicapped vehicle) and 31% returned Survey three (identifying Vehicle B as handicapped vehicle).



Graph 2. Age - Overall mean age was 41-50 years.



Graph 3. Sex.



Graph 4. Vehicle Preference - 20% more respondents preferred vehicle A after it was identified as a "handicapped vehicle." 21% more respondents preferred Vehicle B after it was identified as a handicapped vehicle. Conclusion: If a vehicle accomodates both disabled and non-disabled, then able-bodied populations are more likely to buy it than one intended for use only by the able-bodied.

Statement of Purpose

Public acceptance of this approach to vehicle design demonstrates that design efforts are justified. Our purpose is to develop an affordable automobile for the personal transportation needs of a select group of physically disabled and able-bodied populations.

Design Criteria

The design solution will satisfy the following research based criteria:

- The following populations can enter, start, and operate automobile: able-bodied, paraplegics, triplegics, hemiplegics, quadriplegics, amputees (one arm is needed), obese, short, tall, and pregnant.
- Vehicle accomodates: 2 wheelchair bound and 3 non-wheelchair bound; or 3 wheelchair bound only; or 5 non-wheelchair bound
- $11,000 consumer cost (1985 dollars)
- Maintenance accessible and easily maintained
- Accessible storage space
- Automobile can be entered and exited when parked on street, lot or in a garage.
- Secure against vandalism and theft
- Safe
- Material selection reflects cost, weight, durability and environmental considerations
- May or may not be identified as a disabled persons vehicle
- Clear visibility for driver and passengers.
- Vehicle styling is attractive.

Design Solution

The final solution illustrated and described below is an automobile that can be entered and operated by both able bodied and physically disabled populations. In addition this vehicle satisfies all of the above stated design criteria.

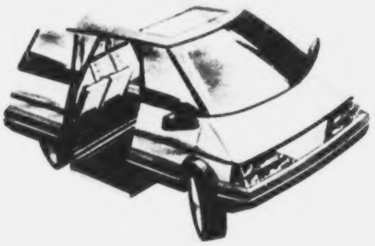
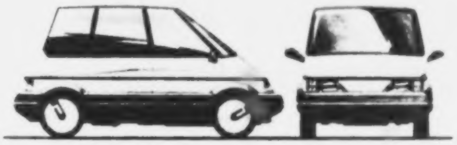


Figure 1. Perspective rendering. This drawing describes overall vehicle form. Electronically activated sliding door (on passenger side only) accesses drop floor. This drop floor electronically raises to vehicle floor level and lowers to street or curb level for wheelchair entrance into or exit from automobile. Spring loaded rear seats automatically fold back for wheelchair mobility.



Figures 2 & 3. Side & Front View Rendering. Windows facilitate clear visibility by all passengers and driver. Window dimension conceals wheelchair when viewed from outside.

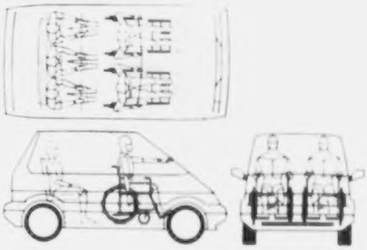
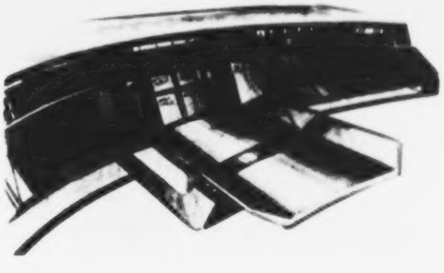


Figure 4. Human Factors. Front, Top & Side views show vehicle with 2 wheelchair bound and 3 non-wheelchair bound inside, OR vehicle will accomodate 2 non-wheelchair bound in front and 3 non-wheelchair bound in rear OR 2 wheelchair bound in front and 1 wheelchair bound in rear.



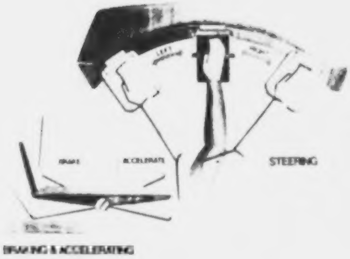


Figure 5 & 6. Control System. Pivoting hand control will enable driver to accurately manipulate steering, accelerator, brake, directionals, horn and other controls with limited movement of one hand. Steering hand pedal is pivoted forward to brake (with finger), and back (with base of hand) to accelerate. Steering is accomplished by sliding hand control left and right along curved dash by pivoting arm at shoulder and elbow. Flush dash panel facilitates cleaning. Controls are positioned on accelerator/brake so that one with limited finger dexterity can accurately access any control with base or side of hand while maintaining full control over steering, braking and acceleration. Control can be set for "zero friction" or "increased friction" depending on limitation of hand strength and movement.

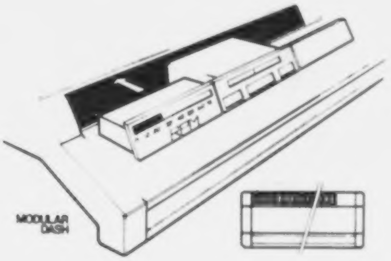


Figure 7. Dash System. Modular dash system allows user flexibility to select and position controls and accessories. Additional dash components may be added, deleted or moved at any time. Cover panel (screen) conceals dash components to reduce the likelihood of vandalism when the vehicle is not in use.

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2.1

TRUNK MUSCLE DIMENSIONS USING ULTRASONIC SCANS: AN OBJECTIVE TOOL FOR LOW BACK PAIN REHABILITATION AND BIOMECHANICS RESEARCH

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ABSTRACT

This paper presents the first use of ultrasound, a safe, noninvasive technique, to measure the dimensions of trunk muscles. Transverse scans of both extensor and flexor groups were taken at the L3 level. Sagittal scans were performed on the extensor muscles at several distances from the midline. Length, width, and area were recorded for the following muscles: rectus abdominis, external oblique, internal oblique, transversus abdominis, and oblique group. Depth of the erector spinae was measured at levels L1 through L4. Error of reproducibility of the scans and measurement error of each dimension were evaluated. Clear delineation of each muscle group was found. Several dimensions, including the rectus abdominis length, the oblique group width, and the erector spinae depth, had acceptably small errors. Results suggest that ultrasound may be a valuable tool for evaluating changes in trunk muscle dimensions.

INTRODUCTION

An important aspect of low back pain rehabilitation, etiology, and functional impairment is the role of the muscles in trunk biomechanics. Two important parameters are muscle size and location relative to the joints that they activate. Such structural information has a variety of applications in understanding trunk and spine biomechanics, as well as the normal and abnormal function of the muscles themselves.

However, because there is presently no good means for obtaining muscle dimensions *in vivo*, research in this area has been hampered. Computerized tomograms are accurate, but involve significant irradiation. A few cadaver measurements have been reported, but significant distortions probably occur during preparation of the specimens. Correlation of these data with measurements in living subjects is not available.

Indirect methods such as trunk strength or EMG (electromyography) may possibly correlate with muscle size. This has been reported for other muscles (1). Such a study of the trunk muscles is presently underway in our lab, but this, of course, requires a methodology for independent assessment of muscle dimensions in order to allow the investigation of correlations.

Ultrasonic scanning for medical diagnostic purposes has developed tremendously in recent years. In particular, "B mode" scanning to define cross-sectional views has seen widespread application. In the past, this has been used to define dimensions of a variety of internal body structures, for example the spinal canal (2), but apparently not muscles.

Reported here is the first application of ultrasound to determine trunk muscle dimensions.

METHOD

Nine healthy male volunteers with no history of low back disorders had cross-sectional scans of the trunk extensor and flexor muscles, using a Picker #80LDI ultrasonic scanner with a 5 MHz transducer on short internal focus (focal zone of 3-5 cm).

Transverse scans were performed of both extensor and flexor groups (Figure 1) at the level of the third lumbar vertebra. Also performed were sagittal scans of the extensor muscles at various distances (0.0, 1.0, 2.0, 3.0, 3.5 cm) from the midline. A set of duplicate scans was taken 10-15 minutes later, with the subject positioning and machine set-up repeated.

Figure 1. Transverse scan of flexor and extensor groups at L3

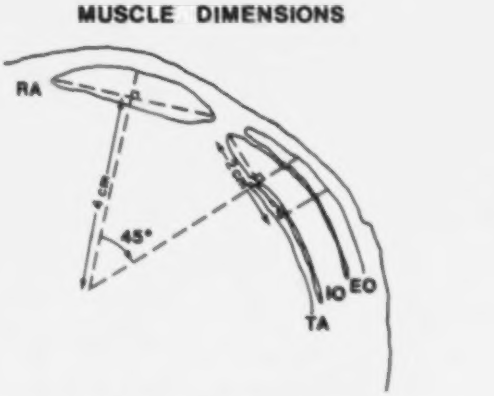

Figure 2. Manual tracing of transverse scan of flexor and extensor groups at L3

The scans were then processed in three steps. First, significant anatomical features were manually traced from the transverse scans of the trunk flexors (Figure 2) and the sagittal scans of the extensors. Second, a local coordinate system was established on the transverse scan of the flexors. The construction lines chosen were used because the scanner available could not provide a large enough scan angle to delineate the posterior margin of the abdominal muscles. Third, various dimensions were measured using a manually-controlled x-y digitizer for the areas and a millimeter ruler for the lengths and widths.

The length, width and area were measured for the rectus abdominis (RA), external oblique (EO), internal oblique (IO), transversus abdominis (TA), and oblique group (OG) at the L3 level. The depth of the erector spinae (ES) was measured on the sagittal scan at the L1 through L4 levels. The scan 3.0 cm off midline was felt to provide the best delineation of the deep surface of this muscle group. The transverse scan of the ES did not provide sufficiently clear detail to be useful in this study.

To assess the reproducibility error with this technique, scans were traced 10 separate times. The first tracing was digitized 10 times and the mean and standard deviation calculated for each dimension (digitizer error). The other nine tracings were digitized and the means and standard deviations calculated for all 10 of the initial digitizings (tracing error). The "combined error" (Table 1) was calculated as the square root of the sum of the squares of these two standard deviations.

RESULTS

Clear delineation of each muscle group is apparent. The mean and standard deviation for the nine subjects for each muscle dimension are shown in Table 1. Variation between individuals is reflected in the relatively large standard deviations for some of the dimensions.

The Error of Measurement reproducibility ("combined error" as defined above) is shown in Table 2, for a typical subject. Of note are the relatively small errors for certain dimensions, such as rectus abdominis length, oblique group width, and erector spinae depth. In contrast, there are other dimensions, such as external oblique area, transversus width and area that have relative errors sufficiently large as to limit future usefulness.

DISCUSSION

An understanding of etiology, treatment and rehabilitation of low back pain must take into account the trunk muscles. Ultrasound scans, presented here for the first time for trunk muscle measurement, allow noninvasive measurement of muscle dimensions, and thus provide the possibility for a series of future investigations using this methodology.

1. Ultrasound scanning is a safe, noninvasive technique which is capable of delineating trunk muscle cross-sectional outlines,

Table 1

Measurement Reproducibility

DIMENSION	VALUE (cm or cm²)	COMBINED ERROR	RELATIVE ERROR (%)
RA Length	7.8	0.14	1.8
RA Width	1.1	0.14	13.0
A Area	4.6	0.85	18.0
EO Width	1.1	0.14	13.0
EO Area	3.1	0.85	28.0
IO Width	1.4	0.14	10.0
IO Area	5.4	0.85	16.0
TA Width	0.6	0.14	25.0
TA Area	2.3	0.85	37.0
OG Width	3.3	0.14	4.2
OG Area	15.4	1.47	9.5
L1 Depth	5.9	0.32	5.4
L2 Depth	5.9	0.32	5.4
L3 Depth	5.9	0.32	5.4
L4 Depth	5.5	0.32	5.8

RA=rectus abdominis; EO=external oblique; IO=internal oblique; TA=transversus abdominis; OG=oblique group.

Table 2

Muscle Size

DIMENSION	MEAN (cm or cm²)	S.D.
RA Length	7.2	0.8
RA Width	1.4	0.6
RA Area	7.0	2.3
EO Width	0.8	0.3
EO Area	3.9	1.5
IO Width	1.1	0.6
IO Area	6.2	0.8
TA Width	0.3	0.3
TA Area	2.3	0.8
OG Width	3.1	0.6
OG Area	20.0	3.9
L1 Depth	5.5	0.9
L2 Depth	5.5	3.2
L3 Depth	5.9	0.9
L4 Depth	5.9	2.3

RA=rectus abdominis; EO=external oblique; IO=internal oblique; TA=transversus abdominis; OG=oblique group.

using routinely available clinical scanners.
2. Measurement of length, width and area of the abdominal trunk flexors, and depth of the trunk extensors has been performed on nine normal human subjects.
3. Inter-individual dimensional variation is relatively large (Table 1) as expected.
4. Measurement reproducibility error is acceptably small for certain important dimensions (Table 2), especially erector spinae depth, oblique group width and rectus abdominis length.
5. Future research is needed to explore the limits of usefulness of this measurement tool. Current projects include correlation to trunk strength, and influence of lumbar corset wear on muscle size.

This research is supported in part by a grant from the National Institute for Handicapped Research and by Camp International, Inc.

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RELATIONSHIP BETWEEN SITTING POSTURES AND LOW BACK PAIN

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ABSTRACT

A preliminary study was undertaken to determine the effects of various sitting postures on the low back. Sitting tolerance was measured for eight different postures. Because of its preliminary nature, this study focused on postures that would be considered extreme. Significant differences were found in the lengths of time that various postures were tolerated. Tolerance times for postures involving various combinations of flexion, lateral bending and rotation were much lower than for the erect neutral position. The length of time for which a posture can be sustained decreases as the center of gravity of the body is displaced further from the neutral position. The most common complaint during testing was low back pain. Further research will attempt to establish limits on how long a particular posture can be maintained without being detrimental to the low back.

INTRODUCTION

The relationship between manual lifting tasks and low back pain (LBP) is well documented. The NIOSH Work Practices Guide for Manual Lifting presents data, derived from a combination of epidemiological, biomechanical, and psychophysical research findings, which establish limits for two-handed lifting tasks (2).

It has been observed (4) that workers sometimes suffer low back pain even when their jobs require little or no lifting. Such workers include drivers, CRT and computer operators, typists, and others who are required to assume static postures for long periods of time. No limits have yet been reported in the literature for static posture tolerances and little information exists for dynamic postures, which include various degrees of flexion, extension, lateral bending, and axial rotation. This is in contrast to the fairly extensive work regarding load-lifting limits such as the NIOSH Work Practices Guide for Manual Lifting (2).

The purpose of this preliminary study is to document tolerance for various sitting postures during certain amounts of forward bending, lateral bending and axial rotation, as measured by subjective assessments of pain. It is hypothesized that posture is as important as the weight of materials handled and that, even in the absence of manual lifting tasks, posture is a significant risk factor for low back pain. Subsequent research will be undertaken in order to establish limits for sitting and standing postures in the workplace and to make recommendations regarding those postures which might minimize risk for developing low back pain.

MATERIALS AND METHODS

Twelve healthy subjects (6 male, 6 female) with no history of low back pain participated in the study. Subjects ranged in age from 19 to 35 years with a mean age of 22. Height ranged from 157 to 193 cm (mean 173 cm) and weight ranged from 45.4 to 95.3 kg (mean 68.2). By random assignment, each subject was tested in each of the eight positions listed in Table 1.

Table 1
Eight Test Conditions

(A) erect (N)
(B) 30° forward flexion (F)
(C) 20° lateral bending (L)
(D) 30° axial rotation (R)
(E) 30° F and 20° L
(F) 30° F and 30° R
(G) 20° L and 30° R
(H) 30° F, 20° L and 30° R.

Each subject was seated on a stool of adjustable height, with no back rest, so that the lower leg made a 90° angle with the thigh. The stool was bolted to a force plate in order to measure each subject's center of gravity. Marks were placed on the stool and force plate at the greater trochanter and coccyx, and an outline of the soles of the feet was drawn in order to position the subject. Center of gravity was monitored throughout the test by a force plate. Position was assessed by measuring the degree to which the trunk deviates from its position in the erect sitting position and was monitored by tactile feedback clamps that did not restrain the subject. These clamps, which simply encompass the shoulder, were first set to a predetermined position which allowed positioning of the subject within 30 seconds. A calibration jig, which mimics the anatomical dimensions of the subject, made it possible to preset the clamps without using the subject as a model.

Electromyographic (EMG) recordings of the erector spinae muscle activity were obtained using two pair of surface electrodes. Each pair was placed 3 cm from each side of the midline at the L3-4 level.

A Visual Analogue Scale (VAS) was used to measure pain. The VAS, a form of cross-modality matching, has been used extensively in the measurement of pain intensity and has been widely reported as both valid and reliable for this purpose (3,5). The scale requires a respondent to place a vertical mark on a 10-cm horizontal line with labelled endpoints. In the present study, the endpoints were labelled "No Pain" and "Excruciating Pain."

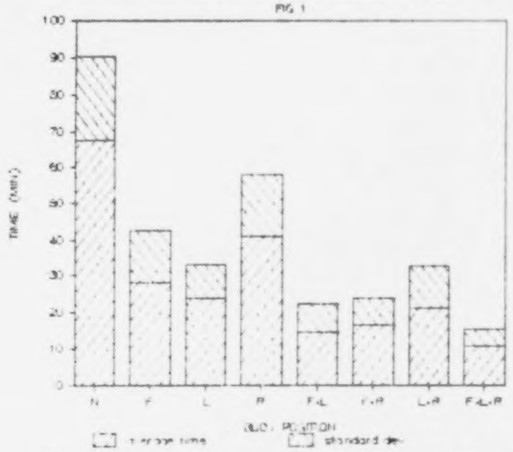
RESULTS

Time to Tolerance of Pain

Figure 1 shows the mean times and standard deviations each of the eight postures was tolerated.

TIME TO PAIN TOLERANCE

FIG 1



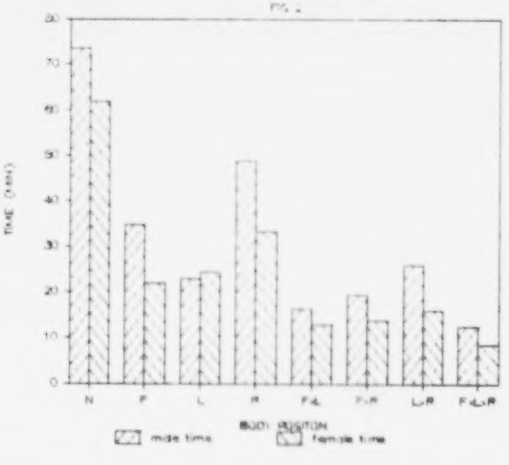
The average time a subject could remain in a neutral sitting position (0° F, 0° L, 0° R) was 67 (S.D. 22) minutes. This was the longest time maintained in any of the eight positions tested. While the neutral position was the most tolerable, the least tolerable position was a combination of F, L and R, with an average time of 10 (S.D. 4) minutes. Subjects were better able to tolerate positions involving only one deviation from the neutral position (e.g., either F or L or R), as opposed to those involving a combination of such deviations.

Rotation was tolerated for an average of 40 (S.D. 17) minutes, compared to 28 (S.D. 15) minutes for forward flexion and 26 (S.D. 11) minutes for lateral bending.

There was considerable variability among subjects in the length of time that they could tolerate a seated posture, as demonstrated by the high standard deviations. Also, males tended to tolerate the various positions longer than did females (Fig. 2).

TIME TO PAIN TOLERANCE

FIG 2



As shown in Table 2, the length of time that the neutral position (N) was tolerated was significantly longer than any of the other postures. The most extreme posture, which combines all three directions (FF, LB, R), was tolerated significantly less than any posture consisting of only one such direction.

Table 2
Probability Levels for Differences in Amount of Time in Various Postures

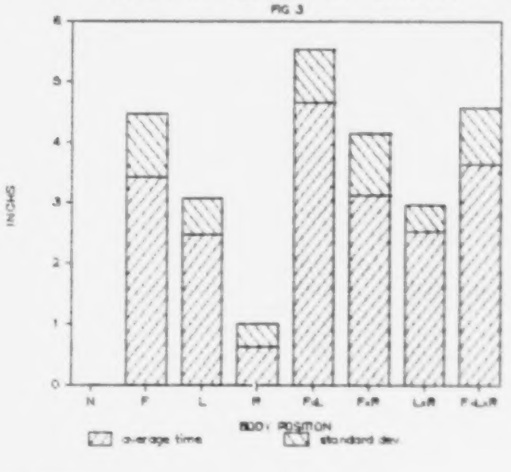
FxLxR	LxR	FxR	FxL	R	L	F	N	
								N
							.05	F
						NS	.05	L
					.05	.05	.05	R
				.05	NS	.05	.05	FxL
			NS	.05	NS	NS	.05	FxR
		NS	NS	.05	NS	NS	.05	LxR
	NS	NS	NS	.05	.05	.05	.05	FxLxR

Center of Gravity

Figure 3 shows the mean shift in the center of gravity for the various postures. As expected, the center of gravity was displaced most in forward flexion, followed by lateral bending. Axial rotation exhibited the least shift in center of gravity. As the center of gravity deviated further from the neutral position, tolerance time decreased, as shown in Figure 3-a.

SHIFT IN CENTER OF GRAVITY

FIG 3



DISCUSSION

These findings indicate that tolerance for seated posture varies with deviations from a neutral erect position. The mean time of 67 minutes for the neutral position does not seem to be very long when compared to the length of time some workers have been observed working in what are essentially static postures. These workers include drivers, typists, VDT operators, and microscopists. Although these workers are free to move, unlike the experimental subjects, they are often constrained to limited motion by various situational demand characteristics such as field of view, control panels, etc. These constraints further exacerbate the problem by sometimes forcing the worker into other than neutral postures.

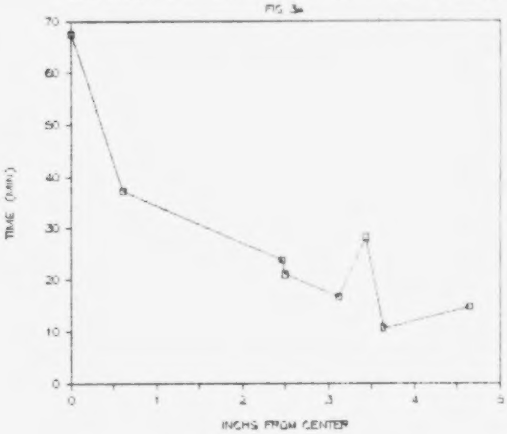

CENTER OF GRAVITY VS.TIME SEATED

FIG 3

While tolerance time changes with deviations from a neutral posture and while a position requiring three such deviations appears most stressful, differences between postures involving two deviations are not clear. Such positions will be further explored in subsequent research to determine relative limits of tolerance. These postures may be tolerated over the short term, but may cause discomfort and low back pain when assumed repeatedly or for long duration during a normal work day.

Chaffin (1) and others have shown that the further from the body a weight is lifted, the more force is generated at the disks and the more tension occurs in the lumbar muscles. A similar phenomenon is seen here, in that the more poorly tolerated postures are those in which the center of gravity has shifted the furthest from resting position.

Interpretation of the EMG data presented some problem in that "noise" (high noise-to-signal ratio) resulted from the low muscle activity. However, in those instances where clear readings were obtained, a frequency shift was observed, suggesting a muscle fatigue phenomenon (6).

Low back pain was the most prevalent complaint of the subjects in all postures. This supports the hypothesis that static postures can cause low back pain. Although the specific cause of the discomfort remains to be addressed, back pain was reported to limit the tolerance of subjects in the present study.

CONCLUSIONS

In spite of considerable inter-subject variability, the present findings suggest that a neutral position can be tolerated longer than postures involving combinations of forward flexion, lateral bending and rotation. As the center of gravity moves further from the neutral position, the time that a posture can be sustained is reduced. Rotation which involves the smallest degree of deviation from the center of gravity was found to be better tolerated than either flexion or lateral bending.

Complaints of low back pain were more prevalent than either buttock pain or upper back pain. Once a subject reported pain, it did not appear to diminish, but rather remained constant or increased. This was true for all 96 tests performed.

While the present study constitutes a preliminary investigation, results suggest a relationship between certain static seated postures and the onset of low back pain. The amount of bending involved in the studied postures was fairly large and may be considered extreme. Further studies will attempt to determine the tolerance for less extreme postures and ultimately to make recommendations for limitations on postures required of workers.

ACKNOWLEDGMENTS

Much thanks must go to Ed Lowe for his help in building the experimental apparatus and to Toni Clark for her editorial and typing assistance.

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2.3

SOURCES OF VARIABILITY IN THE NORMAL TRUNK MUSCLE EMG/FORCE RELATIONSHIP

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ABSTRACT

The function of the lumbar spine extensor muscles was studied by means of the Efficiency of Electrical Activity (EEA) method, whereby the slope of the EMG activity with respect to isometric force during a graded pull is measured. Twenty-nine normal subjects were studied to document 1) the variability with repeated testing and 2) the possible relationship between the EMG/force slope and physical characteristics including age, girth, weight, skinfold thickness, height and sex. The EMG/force relationship was approximately linear, but many subjects had a distinct breakpoint between two linear sections of the graph. Increasing isometric force was associated with greater EMG values than the decreasing force parts of the recordings. There was trend toward decreasing EEA values as tests were repeated. Some significant correlations relating EEA slopes to physical characteristics were found.

INTRODUCTION

A major part of the rehabilitation of many musculoskeletal disorders consists of regimes for the strengthening of muscles. Trunk muscle exercises, in particular, are used in rehabilitation for low back pain. It is not clear (a) whether weak muscles predispose to low back pain or its recurrence, (b) whether low back pain results in disuse atrophy of muscles, or (c), whether strengthening of trunk muscles speeds recovery or helps prevent recurrence (4). Research in this area is hampered by the lack of objective measures of trunk muscle performance.

BACKGROUND

The simplest measure of aggregate muscle performance is the maximum force the subject can exert under various constraints of speed, position and duration. Subject motivation and pain may contaminate these measures. To avoid these potential problems, we have studied the application of an alternative method, the "Efficiency of Electrical Activity" (EEA) to characterize trunk muscle performance. This technique has been used previously to characterize the ratio between the electromyogram (EMG) signal derived from a muscle and the force or moment it generates (1,2). It is especially useful for muscles which have a near linear EMG/force relationship. Maximal force is not required in the measure of EEA. The ratio of EMG to muscle force increases in muscular pathology (5) and decreases with regimented exercise of normal muscle (6).

The work reported here was intended to determine the viability of the EEA as a measure of the integrity of the paraspinal muscles. Prior work indicates that considerable intra-subject variability was to be expected. These muscles do not always have a linear relationship between EMG and force throughout their range of force development. Linearity has been shown in some muscles (1,3), but other muscles with more heterogeneous fiber composition show two or more linear components (7). Cross-talk between nearby muscles could also contribute to nonlinearity. We have opted to measure the first linear portion of the EMG/Force graph, since we were most interested in the low force, pain free range of force development. We studied both the ascending and descending parts of the graphs.

MATERIALS AND METHODS

The experimental technique is illustrated in Figure 1. Subjects stood against a fixture which could provide a reaction force against the thighs and anterior iliac spines while the subjects exerted a steadily increasing effort against a load cell in a cable which connected the harness around the subject's upper trunk to a fixed point in front of the subject. Five Na/NaCl electrodes (a ground electrode over the T-12 spinous process and two bipolar electrode pairs) were attached to the subject's back. The electrodes were 12.5 mm in diameter ("In Vivo Metric") with a 32 mm separation between the centers. Skin preparation involved gentle abrasion with "In Vivo Metric" abrasive swabs. The axis of each electrode pair was aligned in a vertical direction and was 30 mm from the midline of the back at the level of the third lumbar vertebra spinous process. Each bipolar electrode pair was connected to a differential amplifier having an input impedance of 10^6 ohms and a common mode rejection ratio of 70 decibels. The transducer for force measurement was of a strain gauge design and was calibrated to within ±1%. EMG and force signals were recorded on a 'Teac R71' FM cassette tape recorder during the test.

The amplified EMG signal was converted to a DC value by means of an Analog Devices true RMS to DC converter with integration provided by a low-pass, first-order filter having a time constant of 250 msec. These signals were then sampled at the rate of 50 samples per second by the analog to digital converter and the results were stored on disks. Subsequently, graphs were plotted after smoothing the X and Y coordinate data by means of a 25-point (½ second) moving average filter (Figure 2).

Our experience showed that the graphs were either (a) linear, (b) bilinear (two straight-line sections with a breakpoint), or (c) generally curved (Figure 2). A boundary point which corresponds with the breakpoint (D) was defined for each type in order to specify the range of the initial parts of the curves. It was calculated for each curve by fitting a fourth order polynomial to the region between 25% and 75% of maximum and corresponded to the point of maximum curvature (second derivative) in that range.

Analysis of variance and multiple linear regression techniques were used for analysis of the data which consisted of parameters of the EEA curves and anthropomorphic characteristics of each subject (height, weight, age, sex, girth and skinfold thickness at the electrode site).

Figure 1

METHOD OF ASSESSING MUSCLE EFFICIENCY

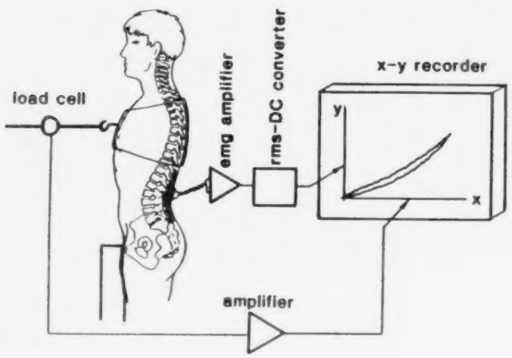
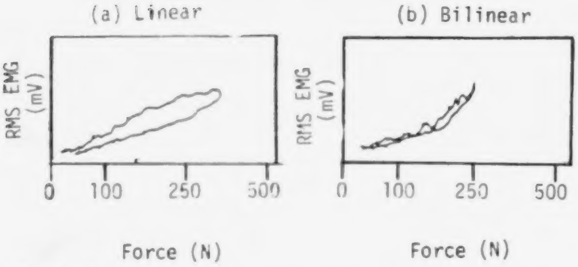


Figure 2

Types of EEA Curves


(a) Linear

(b) Bilinear

Subjects

Twenty-nine subjects were studied. These were healthy volunteers with no history of back pain. The 23 males and 6 females had a mean age of 25 years (range 18 to 40), mean body mass of 70 Kg (range 49 to 91), mean skinfold thickness of 15 mm (range 18-28), and mean height of 176 cm (range 158-198). Eight of the subjects (4 males, 4 females) also participated in a study of intra-subject variability in which they returned for testing five times over two weeks. Each time a subject was tested, two trials were recorded with about one minute recovery time between pulls. The subjects paced each pull to increase muscle force smoothly over ten seconds and then to relax that force completely in ten seconds. Maximal effort was not requested.

RESULTS

The form of the EEA graphs was classified as linear in 14 cases, bilinear with a break-point in 11 cases and generally curved in 3 cases. The mean initial slope was 19.1 mV/kN for the ascending part of the graph (force-increasing) and 15.6 mV/kN for descending (force-decreasing) parts of the graph. Thus, the characteristics of the two parts of the curves were different, producing a "hysteresis" in the graph with increasing force associated with greater EMG values than decreasing force.

There was substantial variation across subjects in the EMG/force characteristics. Using averages of two trials and both placements (left and right) for each of the 29 subjects as a summary figure, the coefficients of variation (SD/Mean x 100) for the ascending and descending slopes were 59% and 44% respectively. Variation for maximal EMG/force and breakpoints in force/EMG ranged similarly from 49% to 55% across subjects.

There was weak correlation ($r = .30$) across subjects between the EEA and either maximal EMG or force, while the ascending and descending EEAs correlated highly ($r=.80$) with each other.

The ascending or descending force slopes were not significantly related to sex, height, skinfold or age of the subjects, although weight correlated somewhat with the ascending EEA ($r=-.36$; $p=.05$) and with descending EEA ($r=-.35$; $p=.06$). Girth correlated to a higher extent with the ascending EEA ($r=-.39$; $p=.04$) and with the descending EEA ($r=-.45$; $p=.01$). The maximal force generated bore a significant relationship jointly to the height (partial $r=.43$; $p=.01$) and the sex ($p=.06$) of the subjects, with males generating the larger forces, on average. The maximal EMG generated was similarly related to the subjects' height ($r=.45$; $p=.01$). The above results were obtained by a multiple regression analysis and backward elimination of insignificant variables.

Based on the 29 normal subjects tested, the following results were also obtained:

1. There was no significant difference in average values between the EEAs obtained in the two trials on the same day, but the average force was higher by 6% on the second pull ($p=.03$).
2. There was no significant difference on average between the EEAs or maximal EMG and force for the left and right electrode placements.
3. The EMG/force characteristics are strongly repeatable from trial to trial (same day) on the same person. The trial-to-trial correlation coefficients for ascending and descending EEAs with left and right placements ranged from $r=.77$ to $r=.86$.
4. The correlation coefficients between left and right EEAs on either of the two pulls were less strongly correlated during the force-increasing pulls ($r=.71$ and $.79$ for the two pulls), and substantially less correlated during the force-release phase ($r=.18$ and $.35$ for the two pulls).

An analysis of variance was performed for the ascending and descending EEAs to assess the relative variability due to subject-to-subject differences (8 normal subjects), day-to-day differences (5 days), left-to-right placement differences, and trial-to-trial differences (two trials per day). In terms of mean differences for the descending EEAs, the left-to-right difference between placements was 11% of the grand mean; trial-to-trial differences within a day averaged nearly 0% of the grand mean. There was a downward trend across the five test days with a change (comparing the fifth day to the average of the first two days) which was 25% of the grand mean. The greatest source of variation was subject-to-subject differences in average slope; the standard deviation among these 8 subject averages was 43% of the grand mean.

DISCUSSION

In this paper, we have sought to find the intra-subject repeatability of the EMG/force relationship in the trunk muscles of normal subjects. Additionally, we have sought to find those anthropomorphic measurements which will explain part of the inter-subject variability.

As shown in Figure 2-b, some of the recordings obtained in this study demonstrated a clear breakpoint between two near-linear sections of the EMG/force graph. A clear breakpoint was found to be present in approximately half of the subjects. It was hypothesized that this breakpoint phenomenon might be due to inherent nonlinearity of the EMG/force relationship in these muscles, possibly because of an abrupt change in recruitment and firing rate characteristics. Alternatively, it is possible that this behavior would result from a macroscopic phenomenon in which additional muscles or regions of muscle were recruited at the force level corresponding to the breakpoint.

There was a consistent difference between the force-increasing part of the recording (the ascending curve) and the force-decreasing part (the descending curve) with the descending part of the curve showing a smaller amount of EMG activity than the ascending part of the curve for the same force in the load cell. As expected, there was no statistically significant difference on average between the activity on the right and the left sides of the subjects. A future aim is to determine whether this inherent symmetry is also present in patients with low back pain.

There was a tendency of the slope of the EEA graph to decrease over time in the subjects tested for repeatability. Since these studies were performed over a period of approximately two weeks, it appears unlikely that this change was due to a real training effect of the muscles. More likely it reflects subjects' greater familiarity with the test, resulting in their relaxation and a consequent reduction in antagonist muscle activity.

CONCLUSIONS

1. There was greater variability in the slope of EEA values between days of testing than between trials in the same testing session, explained in part by the trend of the EEA slopes to become smaller over the days of testing.
2. The gradient of the EEA curve tended to be lower in subjects with greater body weight and girth and the correlations were significant. Adjusting for these two measures would reduce the standard deviation of the EEA across subjects by 9%.
3. EMG activity is less during release of isometric force than with increase of isometric force of the back extensor muscles.
4. In normal subjects, one can expect symmetry between left and right muscle activity during force development, but less during force release.
5. These studies form a baseline from which we can look for differences in subjects with back pain and changes with treatment. The changes in EEA slope over the testing period emphasize the need for control groups when this technique is used to measure changes in muscle function.

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2.4

AUTOMATIC SYSTEM FOR THE EVALUATION OF THE ACTUAL FUNCTIONAL EFFICIENCY OF GAIT

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ABSTRACT

An automatic procedure for the evaluation of locomotor efficiency is proposed. On the basis of the results provided by the Vector Diagram Technique for the analysis of human locomotion, a data collection suitably arranged into homogeneous files is adopted for statistical comparison between different pathologies and normal gait. From the arising differences, computed at two levels of significance and displayed in discrete and continuous forms, indexes of abnormality are proposed to quantify the original deficit and the recovery after specific therapeutic treatments.

INTRODUCTION

Clinical evaluation of the gait efficiency either before or after treatment may be usefully integrated by objective measurements of patient's performances.
The Vector Diagram Technique (VDT) is a particular experimental procedure providing on-line, during the stance-phase of a step, the ground reaction evolution in vectorial form (1, 2).
This technique has been used for testing patients with various pathologies (3, 4, 5, 6) but, although the data so obtained permit immediate preliminary considerations on the motor efficiency of the subject examined, a further mathematical approach is here performed on the VDT results in order to quantify numerically the functional changes induced by a given deficit and, successively, the recovery obtained by treatment. This system is then particularly important to store synthetically periodic controls of the effects induced by rehabilitative exercises, orthoses and surgical interventions.

METHOD

A scheme of the equipment adopted is shown in fig. 1.

Gait analysis is performed by means of two sequential steps:
- acquisition and preliminar process of the ground reaction evolution to obtain normalized vectograms;
- statistical evaluation of the differences between normal and pathological gait.

Acquisition and pre-process

Each patient was asked to walk along a stright pathway at his more natural cadence. Two thirds down, where patients reach steady-state ambulation, a force plate with four cells of piezoelectric transducers was placed flush with the floor. The signals from the platform, which are proportional to the force ratio exerted by the subject on each cell, are firstly amplified and then sent both to a host computer and to a special device, named DIGIVEC (7), displaying on-line the ground reaction evolution in vectorial form. This evolution, usually called vectogram, can be reviewed and qualitatively controlled on a monitor in order to accept or reject the trial. If accepted, the signals sent to the host computer and converted into numerical form at 1 KHz are pre-processed to normalize the data thus making the comparison between different trials possible.
Normalization procedure is:

$$V_n(t_n) = V(t_n) / W$$

$$H_n(t_n) = H(t_n) / W$$

$$X_n(t_n) = (X(t_n) - X_{min}) / (X_{max} - X_{min})$$

where $t_n = t/D$ is the normalized time, D is the stance-phase duration, W is the subject's weight, V, H and X are the vertical and horizontal components and the application point displacement of the ground reaction vector; H and X are computed both in sagittal (HS, XS) and frontal (HF, XF) plane to have a 3D representation.

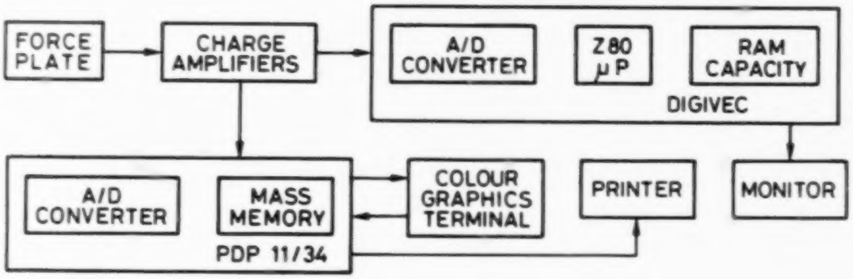

Figure 1

Statistical approach

First of all, the vectograms of normal gait, arranged into homogeneous groups (i.e. including people of similar age, weight, height, the same sex and walking at the same cadence), have been processed obtaining for each group a typical vectogram in which the vertical and horizontal components and the displacement of the application point of the i-th vector appear as mean values and related standard deviations computed from all the i-th vectors of the same group. Similarly, the typical vectogram of a pathological subject has been obtained from five trials.

Since the five variables V, HS, XS, HF and XF of all the i-th vectors concerning either the five trials of a pathological subject or a homogeneous group of normal subjects can be considered as five samples approximately normally distributed aruond their mean values, the significant difference between the normal and abnormal samples has been verified by using the t-test and analyzing the null hypothesis with at 0.01 and 0.05 level of significance.

In order to quantify the differences regarding both each sample (S) and the whole stance-phase (T), the following indexes of abnormality have been computed:

$$S = \sum_{1}^{n} J_{is} \qquad T = \sum_{1}^{m} S_{i}$$

where n and m are the normalized number of vectors common for all the vectograms and the number of variables associated to each vector respectively (in this case m=5).

J_{is} is 1 or 0 depending on the t-test applied to variable S of the i-th vectors is verified or not.

RESULTS

In order to show an application of the method of analysis described above, the results from a male subject affected by foot arch alterations are reported. General morphology and details about the original vectograms from normal gait and this kind of pathology are outside the primary scope of this study but, however, are largely presented and discussed in the references (2, 3) listed below.

Figure 2 shows the results of the t-test applied to all the five variables defining a vector. The horizontal stright lines limit the ranges to accept the t values at 0.01 and 0.05 level of significance. Given the general complexity of this diagram, it is not simple to establish the number of rejected values referring to all the vectors. For this reason the indexes of abnormality discussed above have been computed and the histogram illustrated in fig. 3 has been obtained. As clearly pointed out by the upper graph, significant differences involve mainly the middle part of the stance-phase when the

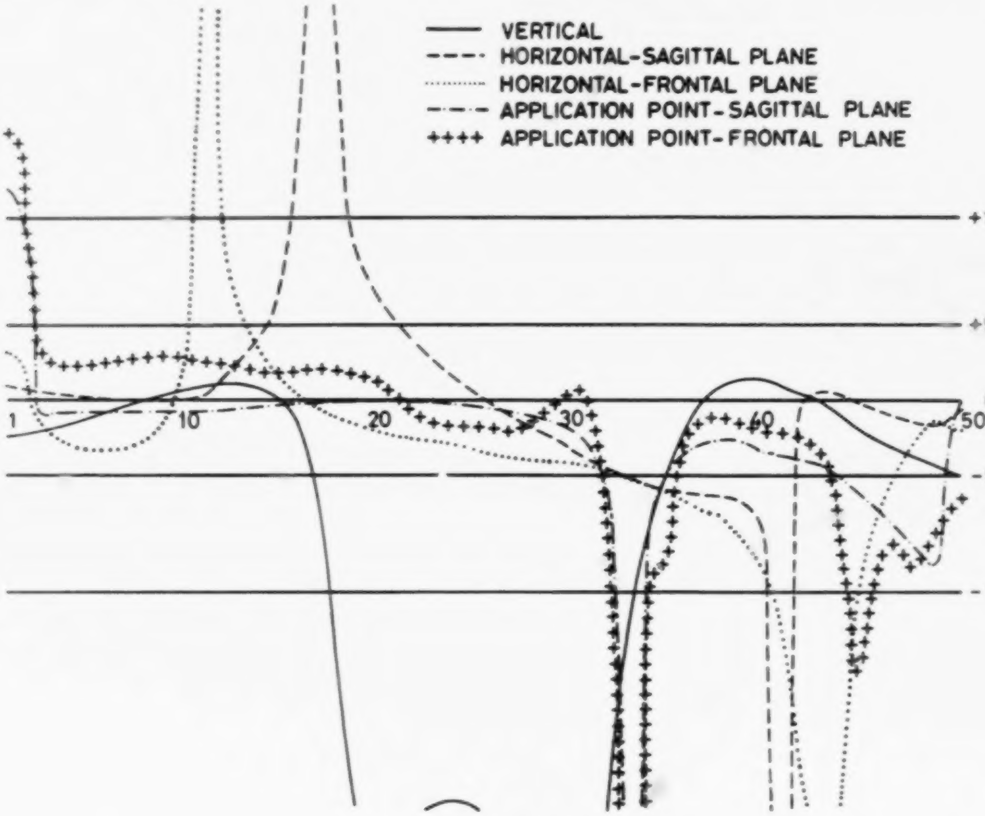

Figure 2

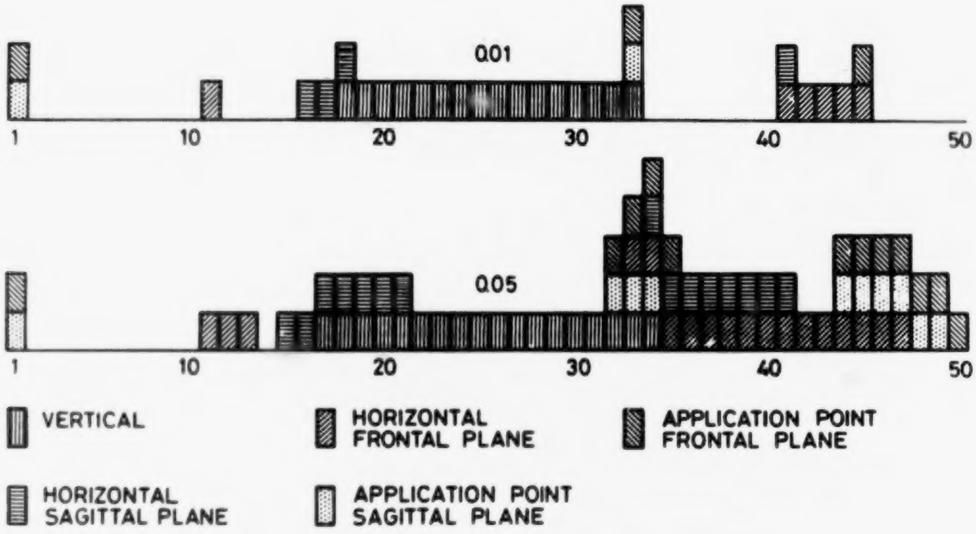

Figure 3

foot plant is fully contacting the ground, that is when the vector is moving towards the fore-foot. Bacause of the pain that takes place medially below the foot, the vertical component cannot be supported correctly thus forcing an abnormal coordination of movement during the following thrust phase as demonstrated by the abnormal distribution shown in the lower graph from the 34-th vector on.

CONCLUSIONS

Although this study is the first attempt to compare statistically all the variables defining the 3-D evolution of the ground reaction vector, the results obtained may be considered much more than merely promising.

The first tests applied to different patients have provided interesting information about the impairment examined and the effect induced on locomotor efficiency.

Further activity will be devoted to improve the reliability of the analysis through additional data collections from normal and pathological gait and a more critical evaluation of the statistical approach.

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2.5

VERMONT SPINAL FIXATOR FOR POSTERIOR THORACOLUMBAR OR LUMBOSACRAL SPINE STABILIZATION: INITIAL MECHANICAL TESTING AND IMPLANTATION

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ABSTRACT

A significant cause of low back pain is mechanical instability of the spine, the treatments for which include surgical fusion of affected vertebrae. In the absence of effective surgical implants to immobilize such unstable segments, the development of a fundamentally new fixation device, the Vermont Spinal Fixator (VSF) has been initiated. To optimize the design of the fixator, various screw design features have been evaluated, varying separately minor diameter, pitch and thread profile. For 6.0 mm (major) diameter screws, maximum pullout strength was attained using a 3.8 mm minor diameter, 2 mm pitch, and a Buttress type thread design. Further investigations underway include evaluation of a 5.0 and 7.0 mm screw, optimal depth of screw penetration and the integrity of the bone-screw interface.

INTRODUCTION

A significant cause of low back pain (LBP) is mechanical instability in the lower back (1). This problem is typically addressed by attempting to obtain a solid surgical fusion of the involved areas. However, there are no optimal surgical implants available at the present time for lumbosacral fusions and none of the implants for the thoracolumbar spine are truly fixation devices which provide immobilization of the unstable segment. Thus, many operations for LBP are performed without using a surgical implant at all, and it is likely that this contributes to the significant rate of failure in achieving a solid fusion (2). A surgical implant which could provide immediate stability and immobilization would substantially increase the likelihood of achieving a solid fusion.

The development and design of a fundamentally new spinal implant, the Vermont Spinal Fixator (VSF) has been initiated to improve stability and immobilization of the thoracolumbar and lumbosacral spine. Compared to presently used implant systems, the VSF offers several advantages:

1. It is truly a "fixator," and does not rely upon compression between implant and bone to achieve stabilization and immobilization (such as is the case with plates and screws).
2. Only very short segments of spine need to be used (e.g., three or even two vertebrae), unlike the Harrington rod system which involves a larger segment of spine, disrupting normal healthy tissue.
3. The device can be readily placed in problem areas such as the lumbosacral region, or laminectomy sites such as the thoracolumbar region, limiting damage to normal tissue.
4. Full three-dimensional adjustability can be easily accomplished, unlike all other implants which afford only limited adjustability.

The VSF (Figures 1 and 2) is attached by means of a specially designed pedicle screw which is placed posteriorly through the pedicle into the vertebral body at the level immediately above and below the site of instability on each side. The two pedicle screws on each side are rigidly joined to a smooth "connecting rod" by means of special clamps. Each clamp allows translational and two rotational movements prior to clamping.

The development phase under discussion here includes the evaluation of screw design features for maximum pullout strength. CT scans of pedicle width, "path length" for screw placement, and pedicle axis orientation were used to determine morphometric constraints on the VSF (Figures 3 and 4). Based on the CT scans, three major diameters for the pedicle screws (5.0, 6.0, and 7.0 mm) were selected. For each major diameter, combinations of various minor diameters, pitches and thread profiles are being investigated, taking into account current metric

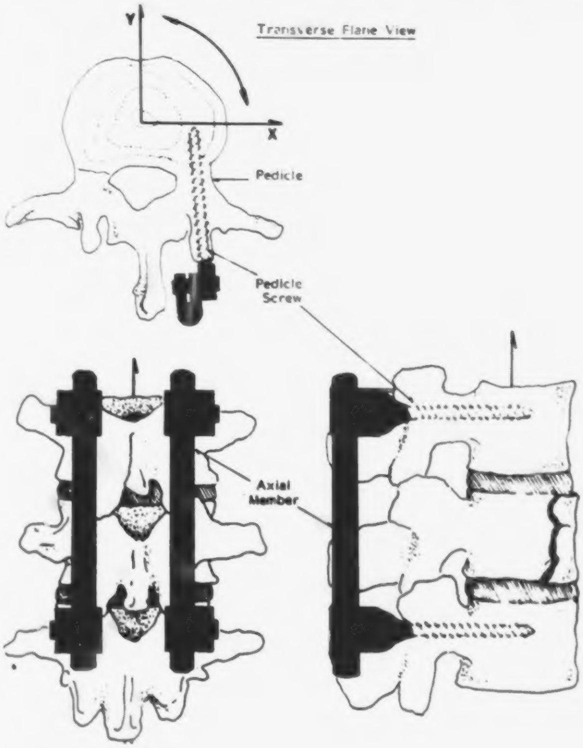

Figure 1. Vermont Spinal Fixator (VSF)

thread standards. Results are presented here for eight different combinations of minor diameter (3.8 vs. 5.0 mm), pitch (2 vs. 3 mm) and thread type ("V"-type vs. Buttress) for the 6.0 mm major diameter screw.

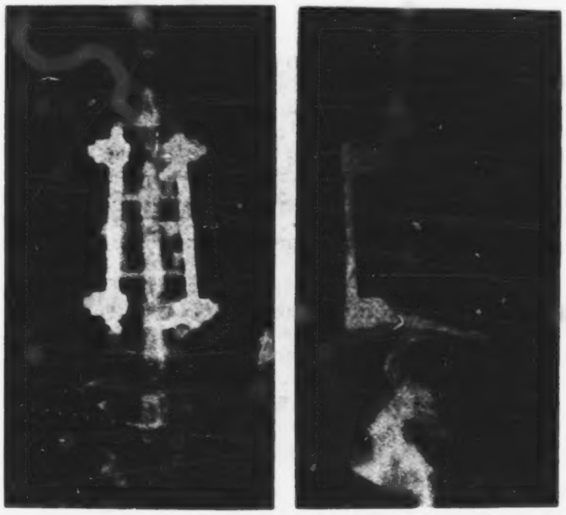

Figure 2. A-P and laberal view of VSF *in vivo*

RESULTS

Ninety-one vertebrae (between T9 and L5) were studied by CT scan to assess critical morphologic features. The mean pedicle width was fairly constant from T9 to T12 (range 6.7 to 7.6 mm) and gradually increased to 14.1 mm at L5 (see Figure 3). Three major diameters for the pedicle screw were chosen (5, 6, and 7 mm). Mean "path length" from posterior cortex of the pars interarticularis to the anterior cortex of the vertebral body gradually decreased from 46.2 mm ±3.7 mm at T9 to 39.9±6.8 mm at L5.

Eight different thread designs for the 6.0 mm major diameter pedicle screw were tested for pullout strength using 32 lower thoracic and lumbar vertebrae to assess the effects of minor diameter (3.8 vs. 5.0 mm), pitch (2 vs. 3 mm) and thread type ("V" vs. Buttress). Left and right pedicles of vertebral levels were considered as experimental units for pullout testing. This was done because of the natural variations in material characteristics of vertebral bodies, both between and within spines. Paired comparisons of strength were obtained by matching different screw types in the left and right pedicles of the various specimens. The levels are thus viewed as statistical "blocks," with the assignment of screw types to left and right pedicle done randomly. Analysis of variance, using a balanced incomplete block design was performed on the resulting data.

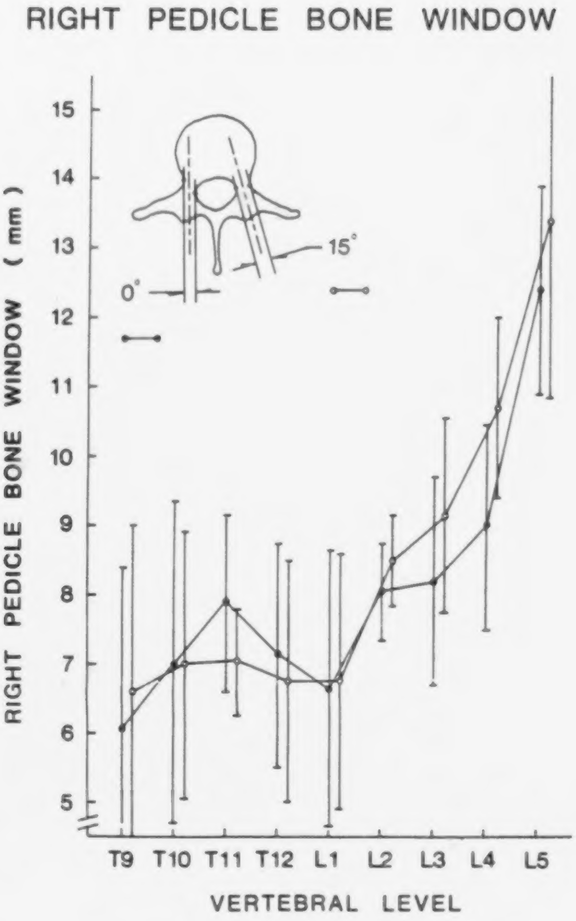

Figure 3. VSF right pedicle bone window

For the "V" profile threads, the smaller minor diameter (3.8 mm) had a significantly greater pullout strength (230 N; $p < .05$) than the larger minor diameter (5.0mm), while pitch (2 vs. 3 mm) had no significant effect on pullout strength. For the "B" profile threads, the smaller minor diameter (3.8 mm) had a significantly greater pullout strength (228 N; $p < .02$) than the larger minor diameter (5.0 mm), and the smaller pitch (2 mm) had a significantly greater strength (242 N; $p < .01$) than the larger (3 mm) pitch. For the "V" vs "B" comparison, initial testing shows an insignificant difference, although further testing is needed to increase confidence limits. Relative mean pullout strengths are show in Table 1. The most promising screw type appears to incorporate the "B" thread type, 3.8 mm diameter, and 2 mm pitch.

Initial cadaver implantation for operative technique and instrumentation development has been accomplished (Figure 2).

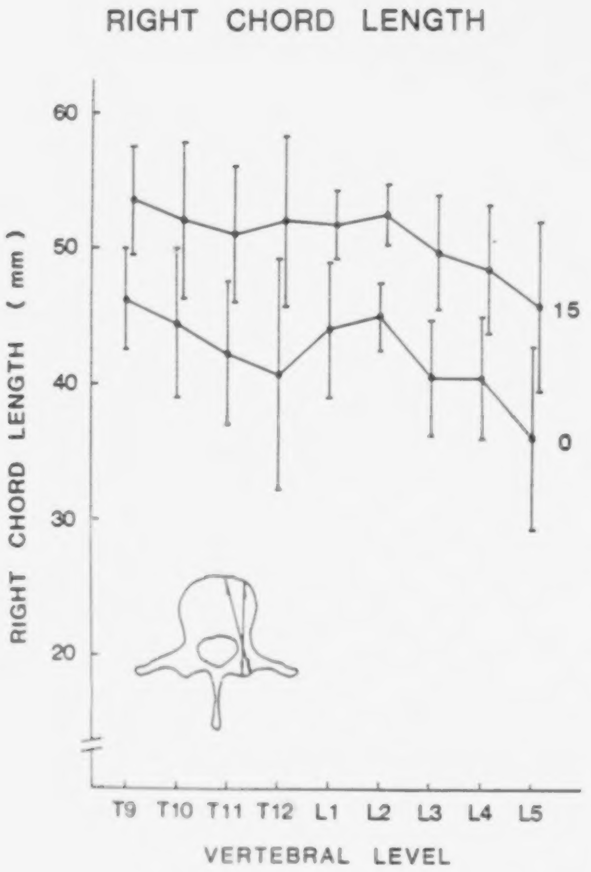

Figure 4. VSF right chord length

Table 1
Mean Pullout Strengths
for a 6.0 mm Pedicle Screw

Mean Relative Pullout Strength	Minor Diameter (in mm)	Pitch (in mm)	Thread Profile*
1.00	3.8	2	B
.83	3.8	3	V
.71	3.8	3	B
.65	3.8	2	V
.59	5.0	3	B
.53	5.0	2	B
.46	5.0	2	V
.34	5.0	3	V

*V="V"-type; B=Buttress

CONCLUSIONS

Eight different thread designs for the 6 mm diameter pedicle screw were tested for pullout strength, using 32 lower thoracic and lumbar vertebrae to assess the importance of thread type. Analysis of variance, using a balanced incomplete block design, showed that, for the "V" thread profile, the 3.8 mm minor diameter had a significantly greater pullout strength than the 5.0 minor diameter, while the 2 mm pitch did not significantly differ in pullout strength from the 3 mm pitch. For the Buttress thread profile, the 3.8 mm minor diameter had a significantly greater pullout strength than the 5.0 mm minor diameter, while the 2 mm pitch had a significantly greater strength over the 3 mm pitch.

Human morphometric data dealing with dimensions of vertebrae relative to screw size and placement have been collected and analyzed. Some of the more critical data, primarily results regarding pedicle width, path length for screw placement and pedicle axis orientation were used to determine morphometric constraints on the VSF. Initial cadaver implantation has been accomplished, with human use planned for the near future. Sizing of the clamps and longitudinal connecting rods to the space available in vivo has been demonstrated to be ample.

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ALTERATIONS OF SKIN BLOOD FLOW PRODUCED BY CYLINDRICAL INDENTORS

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Abstract:

A dimensional analysis is carried out for the alterations of skin blood flow produced by a cylindrical indentor. The associated load distribution and skin displacements are also considered, and displacement measurements are presented for four male subjects.

Introduction:

In the study of skin blood flow as affected by external pressure, a number of different measurement techniques have been employed[1,2,3,4], and skin blood flow is usually plotted against applied pressure. But the use of dimensional quantities such as pressure and flow means that the results must depend upon the subject's weight, size, etc. Furthermore, since interface seating pressures are difficult to measure with accuracy, one often resorts to the use of an indentor for the application of local loading. This means that the results must also depend upon the size and shape of the indentor. Also, since the distribution of loading under the indentor is not uniform, one must either measure that distribution or else make some simplifying assumptions.

Dimensional Analysis:

In order to insure measurement of all pertinent parameters and the proper presentation of results in their most general form, we first perform a dimensional analysis of the problem at hand, which is represented schematically in Figure 1.

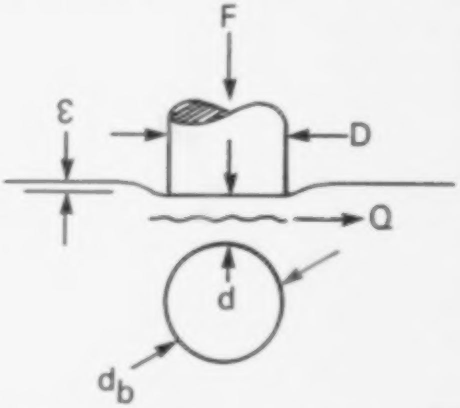

Figure 1. Schematic View of Skin Blood Flow Under Indentor Loading

Here, D represents the diameter of the indentor for application of loads, d is the depth of an underlying bone, ϵ is the deflection of the skin surface caused by the indentor under the applied load F, Q is the volume rate of skin blood flow in the area seen by the blood flow measuring instrument, and d_b is the diameter of the underlying bone (assumed to be spherical or cylindrical). The subscript zero refers to baseline conditions with no load. If we asssume that the flow reaches an equilibrium value for any given load and therefore is not a function of time, we can express skin blood flow Q as an unknown function of the above variables, so that

$$Q = f(F, Q_0, d, \epsilon, d_b, D)$$

But, according to dimensional analysis(5), a physical equation makes sense only if the dimensions on the two sides of the equation are the same. In order to insure that this is the case, we must first represent each of the above variables in terms of its dimensions; e.g., mass, length, and time. Thus, we replace D by L (for length), F by MLT^{-2}, etc. Then, by assuming that the flow rate Q can be expressed as a product of powers of the other variables indicated in the sketch, we find, upon equating the power of each of the dimensions M, L, and T on either side of the equation, that the condition can be satisfied if the ratio Q/Q_0 is a function of the ratios d/D, d_b/D, and ϵ/d.

It is important to note that the force F has dropped out of the above expression and that the variable ϵ accounts for the combined effects of applied force and elastic modulus. In fact, since the elastic behavior of the tissue is most likely nonlinear, the use of actual displacement is preferable to the selection of a single (constant) elastic modulus. Finally, it is found experimentally that, for transverse loading of the proximal femur in side-lying patients, the bone diameter d_b is essentially constant for all subjects. Therefore, the ratio d_b/D is essentially constant, so that the equilibrium flow parameter Q/Q_0 for such experiments is found to depend upon only two nondimensional parameters, ϵ/d and d/D.

Thus, there are three essential quantities to be measured for a proper experiment in the effect of loading on skin blood flow: the blood flow itself corresponding to each loading, the corresponding skin deflection, and the depth of the bone underlying the skin. This last parameter can be measured noninvasively by means of echo doppler equipment.

For a given bone depth ratio d/D, experimental data presented in this manner can be confined within a box bounded by the values 0 and 1 on both axes if we represent the reduction of blood flow in the form:

$$\Delta Q/Q_0 = f(\epsilon/d)$$

That is, blood flow cannot be reduced below zero ($Q/Q_0 = 1$), nor can we compress the tissue more than 100%($\epsilon/d = 1$). Further, there is no reduction of flow for zero displacement (zero load), and there can be no flow when the tissue is fully compressed.

Theoretical load distribution under indentor:

According to reference 6, the load distribution under a flat rigid cylindrical indentor resting on a flat semi-infinite, homogeneous elastic body can be expressed mathematically in the following form:

$$q = (P/2\pi a^2)(1/\sqrt{1-(r/a)^2})$$

where q is the local loading (force per unit area), P is the total load applied to the cylinder, r is the radial distance from the centerline to the point in question, and a is the radius of the cylindrical indentor. It can be seen from the above equation that the pressure at the center of the indentor ($r = 0$) is exactly one half of the average loading over the contact surface, whereas the loading at the outer rim ($r = a$) becomes infinitely large.

The theoretical equation above is of course an approximation to the real situation in several respects. First, it has been assumed that the elastic material (flesh) is homogeneous, and second, the theory does not account for the presence of the underlying bone or for the curvature of the contact surface. However, the approximation is probably a good one for estimating loads at the center of the indentor, particularly since the inhomogoneity of the tissues is primarily in the axial direction normal to the layers of skin, subcutaneous tissue, etc. The general nature of the radial distribution curve seems to be somewhat realistic in that we note a distinct reddening of the skin surface at the location of the outer edge of the indentor, despite a definite machined curvature of the sharp edge there.

Tissue Deformation:

Regardless of the technique to be used to measure skin blood flow, it is a relatively simple matter to measure skin displacement by using marks on the shaft which supports the load to be applied with the indentor. We have used this simple technique to measure skin displacements beneath the indentor with loads applied vertically to the proximal femur of side-lying subjects, and the results are shown in figure 2.

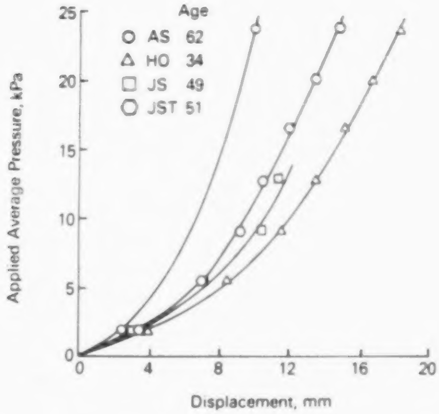

Figure 2. Skin Displacements Measured over the Proximal Femur in Four Male Subjects.

We tested two able-bodied males (A.S. and H.O.) and two paraplegic males (J.S. and J.St.) and found that the "stiffness" of the tissues increased systematically with age, regardless of spinal cord injury. Thus, it can be seen from figure 2 that (1) the elastic behavior of the tissues is in all cases nonlinear, with the tissues getting stiffer (increasing slope) as the load increases and (2) the required load for a given displacement increases with age.

Such displacement information should always be obtained along with any skin blood flow measurements so that the data can be presented in the form $Q/Q_0 = f(\epsilon/d)$ as indicated by the dimensional analysis. This is probably a more meaningful representation than the usual dimensional plot of skin blood flow vs. applied load, for the reasons explained below. However, it should be noted that our displacement measurements were made immediately after each load was applied. This means that transient effects such as creep were not considered.

Discussion:

The reason for performing a dimensional analysis and presenting the results in the nondimensional form presented here is that the results are then in a general form, so that all comparable data should fall on the same curve. This means that, for a correctly selected group of subjects, all data points should fall on the same curve (or surface)[1] Thus, one might expect that all able-bodied subjects would fall on one curve, all upper-neuron lesion spinal cord injury patients would fall on another, all lower-neuron lesion patients on another, etc. unless these categories are inappropriate, or unless other factors are present which have not been accounted for in the analysis.

In any event, if the dimensional analysis has been correctly carried out, and if all of the primary factors have been included, then it should not be necessary to carry out large numbers of experiments to establish the desired curves. Furthermore, the scatter within the various groups should be small.

This procedure is a standard method of engineering analysis and is the basis for all wind-tunnel testing of aircraft. That is, by dimensional analysis, one can use the results of model testing to predict the loads on the actual aircraft without the necessity of building and flying it! It seems reasonable that we should now avail ourselves of this technique in the systematic gathering of experimental data on alterations of skin blood flow produced by external loading, since the fluid and solid mechanics in both cases obey the same laws of dynamic similarity.

It seems from the dimensional analysis carried out herein that skin blood flow measurements are best presented in the form of percentage of flow reduction as a function of the ratios of skin displacement to bone depth and bone depth to diameter of indentor. This would replace the customary presentation of blood flow as a function of applied pressure, since the latter would be expected to depend, among other things, on the weight and obesity of the subject and the elasticity of the tissues being compressed.

Measured skin displacements are presented for four male subjects which indicate that the tissues are elastically nonlinear and that the "stiffness" of the tissues under compression increases with age, regardless of disability. Thus, a higher pressure on the skin indentor is required to produce the same degree of tissue compression in an older subject. It is this compression related to the bone depth (rather than pressure per se) which determines the compromise in skin blood flow produced by external loading. It has been shown herein that such measurements are essential for the proper presentation and interpretation of skin blood flow measurements associated with the application of external pressure.

Acknowledgement:

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3.1

SEATING AND SPINE SUPPORT FOR BOYS WITH DUCHENNE MUSCULAR DYSTROPHY

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ABSTRACT

This paper describes the design, biomechanics, and program being utilized during the past two years for improved seating and spine support for boys with Duchenne Muscular Dystrophy. It relates these factors to quality-of-life issues for the boy and his family.

Perhaps the most certain thing we can say about the development of spine deformity in boys with Duchenne Muscular Dystrophy (including the Becker varient) is that it is highly variable. The pattern may be hyperlordosis, hyperkyphosis, scoliosis, or a combination. While some children will experience an early onset of a spine deformity which progresses very rapidly to a very severe curve, others will acquire only mild, insignificant spine deformity, and still others will appear throughout the spectrum between those two extremes.

There is a lack of precise definitions and numbers regarding the incidence of significant spine deformity in Duchenne M.D., but we can say that a majority (perhaps higher than 85%) of those boys will develop a spine deformity having a large impact on the quality of life for child and parent. Of the deformity patterns we see in this group, those that include large scoliosis curves are the most debilitating. The scoliosis almost always includes a lateral obliquity (tilted laterally) of the pelvis.

The reasons a spine deformity has a negative impact on quality of life are:

1. The trunk progressively loses the stability and posture necessary for independent function of the hands.
2. The sitting position becomes progressively more painful for at least two reasons. Pelvic lateral obliquity causes most of the upper body weight to be borne by only one side of the pelvis. As the scoliotic collapse becomes severe, the lower ribs on the concave side are driven downward against the iliac crest.
3. Distortion of the thorax can eventually compromise heart/lung space and function.
4. As the youngster loses independent hand function and sitting comfort, more burden falls on the family to help with his function and to make constant positioning adjustments for comfort.

It is of fundamental importance, as we treat these kids, that we keep these quality of life issues rather than the orthopedic deformity, per se, uppermost in our minds. For instance, a semi-reclined sitting position will compromise function for these boys more quickly and surely than the spine deformity that position was meant to prevent. The seating and spine support program, whatever it is, must, in a comprehensive sense, help the boy and family to have a greater quality of life.

In the past, virtually all known efforts to orthotically control spine deformity in boys with Duchenne Muscular Dystrophy have failed in the long range. Either they failed outright to achieve the orthopedic goal, they failed to effectively address the quality of life issues, and/or they were excessively technically complicated in their application. At the present time, in some parts of North America, the pendulum is swinging toward early spine fusions.

At Gillette Children's Hospital, we have been seating children with Duchenne (and Becker varient) Muscular Dystrophy for about nine years. Although we were learning much about the nature of the deformity, about the function of these children and their families, and about how to work with them, progress seemed slow until two years ago. In early 1983, we combined the use of a simple soft abdomino-thoracic corset with the Gillette Sitting Support Orthosis (see figure 1). We eliminated all anterior restraints above the lap belt (Previously, we had used vests, panels, or straps to hold the boys back into the supportive plastic shell.). We continued to teach and stress the importance of "pelvic leveling" (elimination or minimizing of pelvic lateral obliquity) each time the child is put into the seat.
We feel this improved seating program has been, thus far, extremely

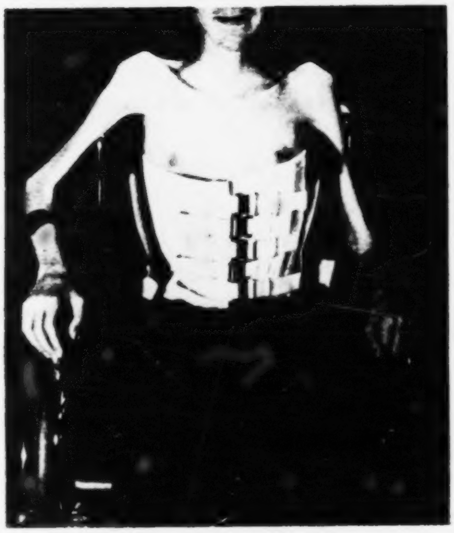

Figure 1

successful. It appears to us that some key biomechancial factors have been addressed in a way that is consistent with improvement of the child's function and other quality-of-life issues.

The normal human torso receives its stability partly from the spinal column acting as a controlled stack of compression elements and partly from a multitude of muscles acting in several different ways. The paraspinal muscles have a direct action on the configuration of the spine; extending it, laterally flexing it, or rotating it. The abdominal (and to some extend the costal) mucles affect the stability and configuration indirectly, but very importantly, through their action on the abdomino-thoracic contents. Muscle action to constrict and control the circumference of the abdomen and thorax allow compressive body weight loads to be taken partly down through the fluid filled abdomino-thoracic cylinder rather than all acting down through the spinal column. This adds tremendously to the stability of the torso. Experiments on a cadaver spine have shown that when a spinal column is stripped of stabilizing musculature, it will buckle (like any flexible elastic column) under a compressive loading of only a few kilograms. As Duchenne Muscular Dystrophy progresses, destroying the effectiveness of the various muscle groups giving stability to the spine, it proceeds to collapse as would any elastic column being acted on by supercritical axial compression loads. The scoliosis pattern we see on x-ray is similar in configuration to the buckling pattern displayed by an elastic column pin joint (allowing tilt) at its top. This biomechanical similarity would be of little practical value except for the fact that the stability of an elastic column under axial compression loads is strongly affected by the nature of the constraint at its end. Constraining the bottom of the column so that it cannot tilt, will roughly double the load carrying capability of the column (see figure 2).

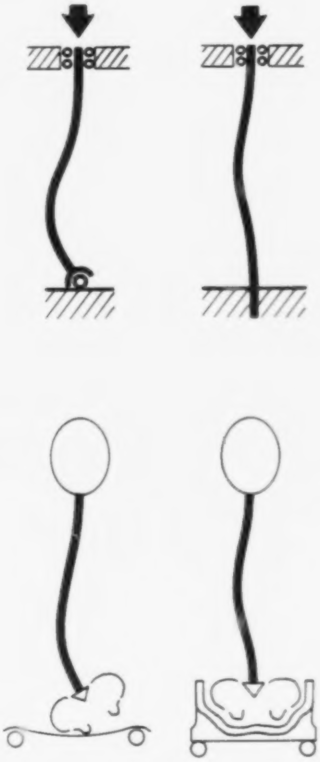

Figure 2

The Sitting Support Orthosis and soft corset combination increase spine/torso stability in three ways.

1.
The soft corset substitutes for non-functional abdominal and costal muscles in a comfortable, unobtrusive way. It allows the compressive loads of body weight to bypass the spinal column by the biomechanism described earlier.

2.
The SSO conforms to the pelvic portion of the anatomy and constrains it to remain in whatever orientation it is positioned in. Whoever puts the youngster into his SSO pulls up on the trousers on the low side of the pelvis and pushes down in the high side to slide the pelvis into a level (or at least minimally oblique) orientation. This is similar to increasing the

bottom end constraint on a flexible column to increase stability. Figures 3 and 4 show x-rays taken just before and after the pelvic leveling procedure was performed on one patient. The Cobb angle is reduced from 36 ° to 20 ° just by this quick procedure performed when positioning the child in his SSO. This patient was not wearing a corset for either x-ray. Please note that once a spine deformity has progressed to a rather severe degree, the pelvis can be leveled only to the degree that the deformity is flexible. We have come, over the years, to see frontal plane pelvic alignment as the most important variable in seating the neuromuscularly impaired youngster with scoliosis.

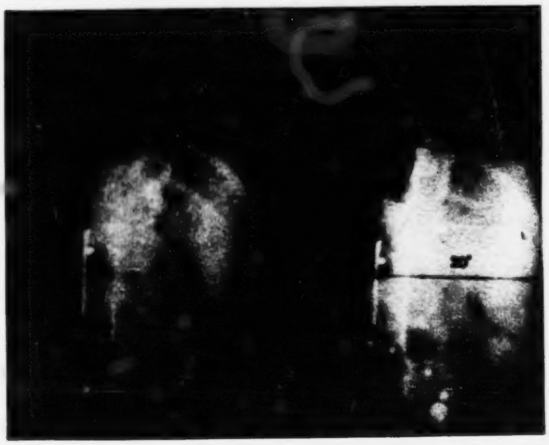

Figure 3 Figure 4

Pelvic control and orientation in the frontal plane relates not only to spine stability and all that derives from that; it relates also to the uniformity of pressures in weight bearing areas and alleviating the progressive deterioration of sitting comfort referred to earlier.

3.
The upper portions of the SSO shell serves as props to prevent the torso, as a unit, from toppling backward, right or left. Forward position of the torso is controlled by the boy per his functional needs.

The corset, when properly adjusted, will also help the diaphragm to have a greater excursion making deeper inspirations possible. Some caution must be exercised. If the corset were adjusted extremely tight, it could compromise pulmonary function by preventing diaphragm descent. It is not difficult to avoid this unlikely complication.

It is the intent of the program here at Gillette to begin the scoliosis control program while the curve is still small (between 20° and 30°). However, some youths enter the program from other clinics with much larger curves. Figures 5 and 6 show same day x-rays with and without the SSO/corset combination (the SSO/corset x-ray includes pelvic leveling). In this case the pelvic lateral obliquity is reduced from 30° to 14° and a 65 ° scoliosis was reduced to 35 °. Curve control of this magnitude is not unusual as long as the deformity is still very flexible.

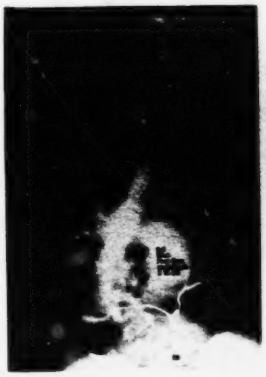

Figure 5

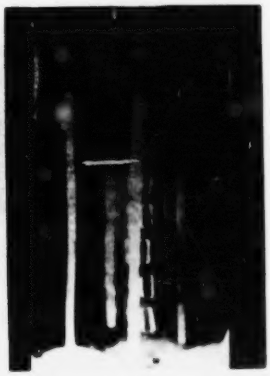

Figure 6

More gratifying than the x-ray evidence is the youngsters reaction. Their comfort, increased sitting height, better sitting stability, and deeper breathing combine to give them a feeling of increased well-being. Their function is maintained or increased, and they feel encouraged.

It must be emphasized that devices do not solve problems of this nature. Only a program including the following things will be successful:

1. Comprehensive education of the family as to the reasons for the prescription and the biomechanical principles.
2. Attention to finding and solving the functional and mobility problems related to seating before they go home with the new equipment.
3. Provide effective equipment and show the family evidence (i.e. x-rays) of its effectiveness.
4. Follow-up to solve growth, function and comfort problems as they occur and to monitor effectiveness.

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HIP ANGLE AND UPPER EXTREMITY MOVEMENT TIME IN CHILDREN WITH CEREBRAL PALSY

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ABSTRACT

This study was designed to evaluate the effect of hip flexion angle on upper extremity function in patients with cerebral palsy. Movement time for shoulder horizontal adduction was measured when the hip was positioned at 50°, 70°, 90° and 110° of hip flexion. It was found that movement time was least at 90 degrees of hip flexion and highest at 50 degrees.

BACKGROUND

The need for proper seating and positioning of children with cerebral palsy have been expressed by several clinicians. It has been observed that proper seating can reduce hyperactive reflexes (1, 2, 4). It has also been demonstrated that seating position affects tonic muscle activity (5, 6, 7). Clinical observations also suggest that correct positioning of the child can enhance neuromuscular control and voluntary function (3, 6). In a case study of an athetoid child (8) and in studies involving nine subjects with cerebral palsy (9), it was found that hand function performance was different when the subjects were placed in seats with different seat surface and backrest angles. However, these differences were not large enough to be statistically significant.

PURPOSE OF STUDY

This study was designed to evaluate the effect of hip flexion angle on upper extremity function.

MATERIALS AND METHODS

Subjects

Subjects used for this study included 10 children who have been diagnosed as having cerebral palsy. None of the subjects had any fixed deformities that prevented the hip from being placed in the positions used for this study. All the subjects were intelligent and could follow instructions. Only those children clinically classified as having mild to moderate involvement were accepted into the study. This was done to avoid the effects of wide variability of muscle tone often seen in patients with cerebral palsy (6). Informed consent was obtained from the parents of each subject according to the procedures approved by the University of Tennessee Institutional Review Board.

Instrumentation

A hydraulic-powered multiadjustable seating device was used to position the children (Figure 1). Hip position was altered by independently changing the angle of the seat surface inclination. An electric goniometer was attached to the left hip joint in order to monitor the position of the joint. Actual position of the hip was provided in degrees through an Analog to digital converter. Upper extremity function was measured using a system of touch-activating switch, an Apple Computer and atelevision monitor. The operating software for this system was developed at the Ontario Crippled Children's Centre. The switch was mounted on a lap tray so that its location was on the same saggital plane as the midline of the body. The lap tray was positioned at the same horizontal level as the xiphisternum. The functional movement consisted of horizontal adduction of the dominant upper extremity from an abducted position to the midline to activate the switch.

Data Collection and Analysis

The subject was positioned in the seating device such that the head, neck, and trunk were in the vertical plane and the knee and ankle at 90°. The head was supported with a neck collar, the trunk with a chest panel, and the pelvis with a lap belt. The hip was then randomly placed at 50°, 70°, 90° and 110° of flexion (Figure 1). Hip flexion angles of 70°, 90° and 110° represented the angle between the planes of the pelvis and femur. However at 50° there was some reversal of the lumbar lordosis. Also at 110° knee flexion was changed to 110° to minimize the tendency of the subject to fall forward. A rest period of 10 minutes after each hip position change allowed the subject to adapt to the new position.

The subjects were instructed to move the arm from the starting position (shoulder abduction) to the midline (shoulder adduction) to touch the switch and withdraw the hand as quickly as possible on cue from the television monitor. The movement was repeated 10 times. The 10 movement times (the time interval between the cue to move the arm and activation of the switch) were averaged and represented functional performance for the particular hip flexion position. The differences between the mean movement time obtained for each hip flexion was tested for statistical significance using an analysis of variance and Fisher's Multiple Comparisons.

RESULTS

The mean upper extremity movement time obtained with the hip at 50° was 1.87 seconds (Table 1). Mean movement times of 1.52 seconds, 1.07 seconds, and 1.48 seconds were obtained at the 70°, 90° and 110° hip positions respectively. As shown in Figure 2, movement time was highest at the 50° hip flexion position and lowest at the 90° position. The results of analysis of variance showed that a significant difference existed between the means of movement time obtained at the different hip flexion positions at the 0.05 level. However, the Fisher's Multiple Comparisons indicated that the significant difference was between the 90° position and the 50°, 70°, and 110° positions. The difference between the 50° position and the 70°and 110° position approached a significant level (0.06). The means at 70° and 110° were not significantly different.

DISCUSSION

The results of this study indicate that the hip flexion angle affects upper extremity function as measured by movement time. For the type of patients and at the upright body orientation used for this study, functional performance was highest when the hip was positioned at 90° of flexion. At the 50° position it was observed that breathing was much more labored perhaps due to the decrease in the thoracic and abdominal cavities and may have affected functional performance. It is being recommended that the hips be positioned at 90° when seating children with cerebral palsy with mild to moderate involvement for upper extremity function.

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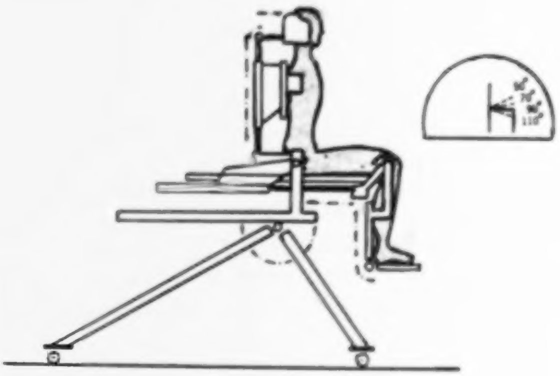

Figure 1 - Side View of Positioning Device. Hip Flexion Angles used are shown on insert on right.

Table 1 - Means and Standard Deviation of the mean for Movement Time for the 4 hip flexion angles.

Hip Flexion Angle (Degrees)	N	MEAN	STD. DEV.
50	10	1.87	0.633
70	10	1.52	0.312
90	10	1.07	0.250
110	10	1.48	0.316

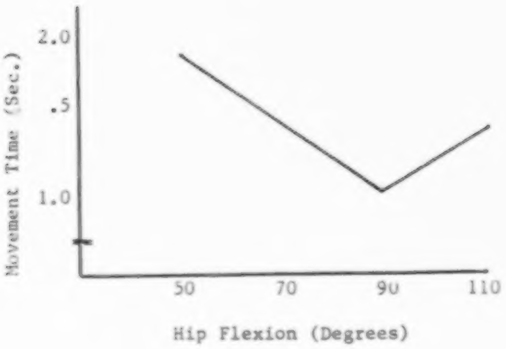

Figure 2 - Mean Movement Time Plotted against hip flexion angle.

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3.3

PRESSURE RELIEF MONITORING DEVICE

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ABSTRACT

A device was developed to evaluate a wheelchair user's pressure relief performance. This device operates continuously without intervention for several weeks and allows the subject to follow normal routines outside the institutional setting. The information collected by this device serves in training the client to perform adequate pressure relief as part of a comprehensive program to prevent tissue trauma.

INTRODUCTION

Wheelchair users with sensory losses are at risk of developing pressure sores because they sit for long periods of time and subject their tissues to pressures which restrict blood flow. Research by Kosiak (1) has shown an inverse relationship between the magnitude of the pressure and the length of time at that pressure which causes tissue trauma. Thus, much of the research and development in seating has been directed towards designing wheelchair cushions which lower the pressure on the user's skin. This work, while exemplary, has yet to lower the pressure to a point where the user may sit without pressure relief for long periods of time.

Problem

It is therefore necessary for wheelchair users with sensory loss to learn pressure relief techniques and practice them regularly in order to prevent tissue trauma. Rehabilitation professionals (RP) teach their clients how to perform pushups (push oneself completely off the seat), side leans and front to back rocking. The instructor and client develop a strategy for pressure relief at regular intervals (e.g. once every twenty minutes) and of sufficient duration (at least five seconds) to prevent tissue trauma. This learning process lacks, however, a method of evaluating the client's compliance with this regimen.

Recent work by others has developed evaluation tools which aid this learning process. Fisher and Patterson (2) monitored pressures beneath the ischial tuberosities of tetraplegic patients using two pressure transducers on each tuberosity. The devices were attached to a 24-hour tape recorder, monitoring on average 5 hours per session, and were replayed through a computer. Chawla et al. (3) developed a training and monitoring device incorporating a pressure sensing pad placed on the sling seat of the chair with a large apparatus placed underneath the seat. This system provided audible and visual cues for pressure relief and recorded when the cue was given and how often pressure relief was undertaken. Mervitz et al. (4) developed a pressure relief monitor using a portable computer. This device runs for two days before its batteries need replacing. Barbenel et al. (5) modified a digital watch to record the ratio between time spent sitting and time spent pressure relieving. This device failed to detect more than one type of pressure relief (pushup) and did not record the time or number of events. The authors are also aware of a monitor/alarm device being developed by Andrews (University of Strathclyde) which utilizes pressure switches to detect pushups.

Approach

To further this work and assist the pressure relief training process, the authors have developed a device which will evaluate pressure relief performance over a time span of several weeks. RPs can use this device with ease because it requires no special knowledge of electronics. Also, the device requires no intervention during the evaluation period. In addition, the subject can follow a normal routine in his every day environment while undergoing this evaluation. The information gathered by this device shows the client and therapist the complete history (type of pushup, time of day executed, and duration) of his pressure relief behavior. With this information, the client and therapist can see when pressure relief performance falls below acceptable levels. They can then develop new strategies to improve performance, including more intensive training and possibly, the use of warning devices (Flower, 1984).

Overall Device Description

The device attaches to the under side of the subject's wheelchair cushion as shown in figure 1. The wheelchair cushion cover envelops the device-cushion combination to give the appearance of a single unit. The device is designed to fit a standard adult wheelchair. Another version of the device fits a narrow adult wheelchair. The dimensions of the device are such that the subject can perform the same activities while undergoing an evaluation as he can with his normal wheelchair cushion. The electronics inside the device are powered with three six volt lithium batteries. A waterproof insulated case encloses all of the electronic components.

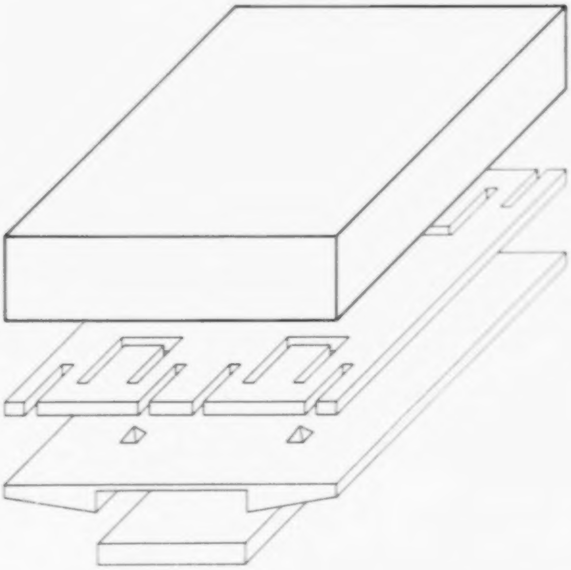

Figure 1. An exploded view of the device which shows (from top to bottom) wheelchair cushion, force plate, four point support system and monitoring box.

DEVICE OPERATION

To activate the device, the RP first sets switches on the device to indicate the client's physical characteristics and the type of wheelchair cushion being used. Next, the client and the RP calibrate the device. The client sits on the device cushion combination. When the device beeps, it ques the client to perform the pressure relief activities within his capabilities. At the same time, the RP monitors the pressure under the client to make sure he is performing effective pressure relief. Once the calibration is completed, the device has set pressure relief event thresholds for the client under evaluation. At this point, the device begins monitoring the client's pressure relief activity for 30 days without intervention. At the end of the evaluation period, the RP connects the device to a personal computer. A software package reads the information collected by the device and stores it on disk. The RP and client can then view the information on the screen or obtain a graphical printout. The display software also plots statistical information about the pressure relief activity such as running average time between events, cross correlation between event frequency and time of day and frequency distribution of event duration.

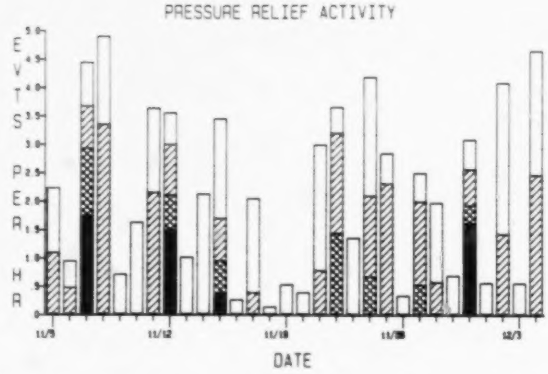

Figure 2. Simulated printout of the performance data. Note the three different types of events.

DESIGN

The device consists of three major components:

1) force transducer
2) detection circuitry
3) monitoring

The first two components measure the spatial coordinates of the center of force on the plate with the client sitting on the wheelchair cushion. When the client makes a body movement, the coordinates of the center of force changes, and if the magnitude of the movement exceeds the preset thresholds, the detection circuitry turns on the monitoring computer. This component then determines if the client is performing a valid pressure relief activity by comparing with the calibration recorded at the start of the evaluation. If a valid pushup has occurred, the computer records the type, time and duration of the event in nonvolital memory. After 20 seconds have elapsed with a valid event occurring, the monitoring computer turns itself off, helping to conserve the battery charge.

Monitoring Computer

The capabilities of the monitoring computer include:

8 bit Z-80 CMOS CPU 500Hz
16K bytes of nonvolital data memory
4K bytes of program memory
real time day/date clock
audible alarm
two 8 bit digital input-output channels
eight single ended 8 bit analog input channels

The developers based the monitoring computer on the Z-80 CMOS CPU so that they could draw on the development software available for the Z-80 and at the same time build a device which requires little operating current. The relatively slow clock speed of the CPU (500KHz) gives adequate performance while reducing the overall power requirements. The 16K bytes of data memory is large enough to hold information for more than a month of valid events based on twelve hours of sitting per day and six valid events per hour. A separate battery continuously powers the data memory and the real time clock for more than a year before it needs replacing. The audible alarm serves to cue the client during the calibration procedure and to signal a malfunction during an evaluation. The digital input-output channels read the switches set by the RP to indicate the client's physical characteristics. These channels also communicate with the display software in the personal computer via a parallel to serial conversion process. The analog input channels read the electrical signals from the force transducer and associated circuitry.

Force Transducer

The force transducer is an aluminum force plate with four integral load cells. The plate makes contact with the wheelchair at only four points, one under each load cell. Each load cell has four semiconductor strain gauges in a full bridge configuration. A simple circuit uses the four strain gauges as feedback resistors on operational amplifiers. This low power circiut is powered up continuously during the evaluation and converts changes in force on a load cell into voltage changes which can be used by the detection circuitry and the monitoring computer.

Detection Circuitry

This circuitry compares the output from the monitoring computer with preset thresholds. When an output exceeds a threshold for more than one second, the circuit turns on the monitoring computer by latching a transistor switch. When the monitoring computer signals turn off, the detection circuitry unlatches the transistor switch. Both the load cell circuitry and the detection circuitry are on continuously. The current required by these two circuits is so low that they can operate for several months before their battery runs down.

SOFTWARE

The above circuitry requires software to operate. This software is composed of three modules:

1) event recording
2) communication
3) display

The authors used Z-80 assembly to write the event recording programs and the C language to write the communication software. Presently, the display software written in BASIC runs on an IBM PC. Versions for Apple, Tandy and other personal computers will be developed as needed.

Event Recording

This software resides in the monitoring computer and processes the pressure relief performance data as it is collected. This software reads the forces on the force transducer over the analog input channel. By calculating the change in the force vector, the software can differentiate between valid and nonvalid events. The software ignores the latter to save data storage space. If a valid event is detected, the software notes the time at which the event began and then waits for the event to finish. When the event is over, the software calculates its duration and codes it along with the type of event and the time of occurrence. This information is then stored at the next available memory location. If a valid event lasts for more than a minute, then this event is regarded as an end of a seating session. The software records the beginning and end of each seating session. After twenty seconds have elapsed without a valid event occurring, the software stores the time and the next available memory location and signals the detection circuitry to turn off the monitoring computer.

Communication

This software resides in both the monitoring and personal computers. When the two computers are connected by the serial port (RS232) on the personal computer and the parallel input-output channel on the monitoring computer, the two computers can exchange data. The software in the monitoring computer converts parallel data into serial data and vice versa when sending and receiving data. The software in the personal computer establishes the communication link and requests the monitoring computer to send a packet (128 bytes) of data. The personal computer performs error checking on the received data by adding up the bytes and comparing the sum with the sum sent by the monitoring computer. If the comparison fails, the personal computer requests that the packet be sent again. When error free data is received, the personal computer stores it on the disk.

Display

This software serves as the executive of the data transfer, analysis and display process. The RP simply places the disk with this software in the personal computer and chooses from the options displayed at the screen. The RP indicates when the monitoring computer has been connected to the personal computer and enters the client's name and hospital number. The software then reads the data from the monitoring computer and displays the dates for which the data was collected and the monitor identification number. Next, the software displays a summary graph of the data by plotting a histogram of the number of valid events for each day. Each bar is divided into sections to indicate the type of event performed. This plot also indicates the mean time between events and the mean duration of events. At this point, the RP can choose from a menu to print out the histogram or to view other plots of the data.

SUMMARY

The need for a better evaluation of a client's pressure relief performance prompted the authors to develop the performance evaluator described above. Currently, the system is in the form of a clinical prototype, but will be introduced routinely into the pressure sore prevention management program at Helen Hayes Hospital in the near future.

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THE SUBASIS BAR: AN EFFECTIVE APPROACH TO PELVIC STABILIZATION IN SEATED POSITIONING

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ABSTRACT

The components commonly designed to control posterior pelvic tilt (PPT) and increased extensor tone causing "pelvic thrusting" are often inadequate in insuring appropriate, consistent, comfortable, and objectively reproducable seated positioning. The authors have developed a system designed to apply a posterior/inferior force, inferior to the anterior superior iliac spine. The direction and point of application of this force effectively reduces PPT and virtually eliminates positioning problems caused by increased extensor tone at the hips.

INTRODUCTION

One of the most difficult, and universal problems facing the professionals involved in the seated positioning of non-ambulatory individuals is posterior pelvic tilt. For purposes of this presentation, posterior pelvic tilt (PPT) is measured with the client/patient in a seated position with hips in 90 degrees of flexion, or to the degree of flexion allowed by passive range of motion. Using this position as neutral; PPT is described as a deviation in which the posterior superior iliac spines move posteriorly and inferiorly.

NOTE: In describing and measuring flexion angle at the hips for purposes of determining seat-to-back angle in the design of seated positioning orthoses; it is critical to differentiate between true hip flexion occuring in the acetabulum, and apparent hip flexion occurring at the lumbosacral junction or even more superiorly in the lumbar region of the spine.

BACKGROUND

Etiology

The etiology of PPT are myriad. In our experience many cases of PPT are unfortunately caused by, or at the very least exacerbated by, inept attempts at providing seated positioning equipment. A common example of this is illustrated in Figure 1. To successfully reduce PPT it is essential to apply a posteriorly directed force inferior to the anterior superior iliac spine (ASIS). If the force is applied superior to the ASIS, it will actually increase the problem, and increase the attendant spinal deformities.

Increased extensor tone at the hips

A second area of concern in dealing with pelvic stabilization is much more active and violent. It is a situation typified by "pelvic thrusting" and usually caused by increased extensor tone at the hips. There is usually minimal structural or functional deformity, probably due to the fact that it is an intermitent phenomenon. The thrusting may be symetrical in the medial/lateral plane, or it may include a rotational component.

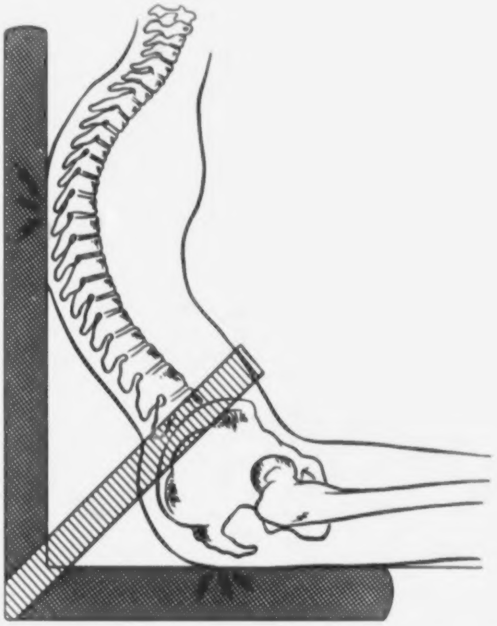

FIGURE 1

One key factor in the success of designing appropriate seated positioning systems for individuals with this type of involvement is to match the seat-to-back angle of the seating system to the limitations (if any) of actual acetabular hip flexion available. It seems almost universal practice in dealing with increased extensor tone, to use a seat to back angle anywhere from 85 degrees to an extreme of 60 degrees. This is often done without regard to actual hip flexion available or to potential effect on viscera and internal organs. This approach may be successful in maintaining an individual in a seating system, but will almost inevitably place the pelvis in posterior tilt, introduce a "C" type kyphotic curve, and have a deleterious affect on head control.

We feel that the more rational and appropriate approach to this problem is a combination of a more conservative seat-to-back angle (approximately 85 degrees), 5 to 10 degrees tilt of the entire seated positioning system, and an appropriate pelvic stabilization component to generate a posteriorly/inferiorly directed force applied inferior to the ASIS.

Pelvic stabilizers
There are a number of "pelvic stabilizers" in common use. We are not attempting to generate a comprehensive listing of all pelvic stabilization components. The three most common types of components are: the "seat" belt, peroneal (groin) straps, and the "pommel".

"Seat" belts. The seat belt can be an extremely effective tool if used correctly. If the seat belt is attached so that the line of pull is below the ASIS in a posterior/inferior direction and it is securely fastened it will ususally be successful in maintaining pelvic alignment. The major failing of the seat belt is that it`s appropriate use is entirely subjective. How tight is "securely fastened"? Even a loosely fastened seat belt will keep the individual in his wheelchair, but not necessarily appropriately positioned. If the seat belt is loosely fastened, and the person is allowed to "thrust" against it repeatedly, the shear force generated will evevntually cause discomfort and the potential for breakdown. Because of the inherent flexibility of a webbing belt, the seat belt conforms to the rounded contours of the pelvis and is usually ineffective in dealing with pelvic rotation.

Peroneal straps. Because they apply a posteriorly directed force lateral to midline, and usually well inferior to the ASIS, peroneal straps are effective in dealing with pelvic rotation. There is the same subjectivity involved in the proper securing of the peroneal straps as is involved in proper use of a "seat" belt. The additional problem of acceptance of "groin" straps, especially by older client/patients, is a concern.

"Pommels". The third, commonly used "pelvic stabilizer", is the "pommel". We firmly believe that the groin is not an appropriate weight bearing area. There is the obvious problem of the external genitalia. Applying force to the area of the symphisis pubis may be somewhat effective in limiting the amount that the patient/client can slide forward as a result of PPT and/or "thrusting. It does not, however, effectively reduce pelvic tilt, nor does it have an effect on pelvic rotation. We feel, however, that the use of a well designed "pommel" as an abduction wedge to reduce adduction contracture and to maintain hip alignment is appropriate.

SUBASIS BAR

To adequately address the need for reduction of posterior pelvic tilt, to maintain pelvic stability in the case of increased extensor tone, to prevent pelvic rotation, and to provide for consistent, objective repositioning by various primary care givers, we have developed the subasis bar.

Design. The subasis bar is a round, padded aluminum bar which attaches to the lateral hip pads of a seated positioning system. The diameter of the aluminum bar ranges from 1/4" to 3/4", depending on the size of the patient/client. The bar is covered with 1/2" of pelite; a dense polyurethane foam. It is inserted in a slot in one lateral hip pad and then secured into a spring loaded bracket on the opposite lateral hip pad. It is designed to be fastened and removed using one hand. This allows for maintaining appropriate pelvic alignment while inserting the subasis bar. Consistent,reproducable,objective repositioning is assured because the subasis bar mounting system allows for only one appropriate position.

Location. The subasis bar is carefully positioned to fit in the area of the femoral (Scarpus) triangle (see Figure 2). This is the anatomical area bordered by the sartorius, the inguinal ligament, and the adductor tendon. This area, especially during sitting when the adductor tendon is relaxed, provides a viable area to apply force. This is the same area that is utilized by the prosthetist for suspension of the quadralateral socket of a prosthesis for an above-knee amputee.

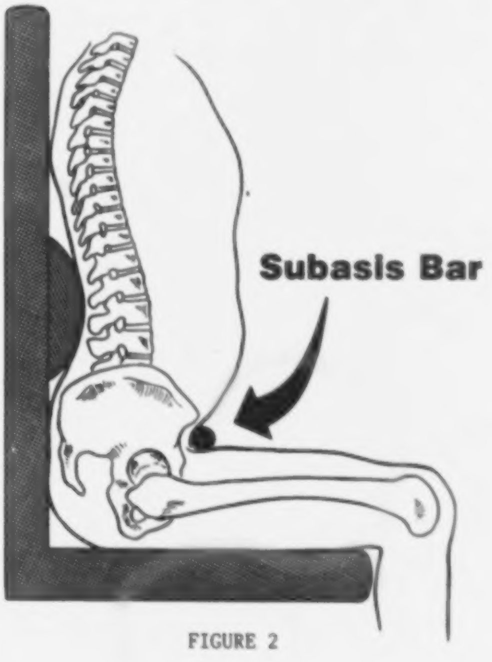

FIGURE 2

Clinical parameters. To be effective, the subasis bar must be used in conjunction with a well designed seated positioning system including a minimum of a firm padded seat and back, and lateral hip pads. The client/patient must be positioned so that the sacrum is perpendicular to the seating surface and parallel to the back surface. Hip flexion to 90 degrees is desirable; pelvic tilt reduced to neutral is optimal, though we have had some success with moderate, fixed deformities. In the seated position, with the hips flexed to 90 degrees, there must be at least 3/4" clearance between the inferior surface of the ASIS and the shaft of the femur.

Clinical experience

The subasis bar has been used in over 50 cases. In all cases, posterior pelvic tilt was reduced to neutral; as allowed by passive range of motion. Problems with maintaining appropriate position due to "pelvic thrusting" have been virtually eliminated. Positioning of the subasis bar inferior to the anterior superior iliac spine eliminates the potential for breakdown over bony prominences. Patient/client and primary care giver acceptance and compliance is assured through careful explanation of the rationale behind the design of the subasis bar.

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A LEVER DRIVE SYSTEM FOR WHEELCHAIRS

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ABSTRACT

The various stages of development of a lever drive system are described. The use of both roller and friction clutches are discussed and the means for controlling forward, reverse and braking are included. The current system allows good maneuverability without requiring hand skills and may be effective for quadriplegics.

INTRODUCTION

Lever drives for wheelchairs have been in common use in Europe for many years. The typical design uses connecting rods from the lever to the drive wheel similar to that used on a steam locomotive. Steering is accomplished by turning a handle on the lever, the handle being connected to a front caster. Bennedik, Engel and Hildebrandt[1] established that this is a more efficient method for propelling a wheelchair compared to hand rims. However there are some associated disadvantages. Top speed is limited by the maximum stroking rate and it is not possible to rest between strokes since the lever is always connected to the drive wheels. Also maneuvering is much more difficult in close spaces since castering action is not possible and a lever might have to be started at a position of dead center. This last feature contributes to the difficulty in mounting a ramp from a standing start.

One of the reasons for increased efficiency probably stems from the fact that power is applied while pushing and pulling. However Brubaker[2] established that using only the push stroke, levers could be more efficient than hand rims, particularly if the appropriate mechanical advantage was used. To exploit this possible increase in efficiency, Bruning[3] designed a lever system that used a roller clutch to drive a sprocket on the drive wheel like a bicycle. This allowed pausing between strokes as well as choosing the length of stroke. Reverse was achieved by designing a reversing roller clutch which was controlled by pressure of the palm of the hand for forward motion and by finger pressure for reverse. Neutral or free wheeling occurred when no pressure was applied. The concept proved to be quite effective but the only braking possible was by engaging reverse, which locked the drive wheels. This did not allow speed and directional control while descending slopes. Caliper, bicycle style, brakes were tried with the hand levers mounted on the main levers. This required three hand functions for controlling forward, reverse and braking, which proved too much for fast action, and it was not uncommon to strike a wall or other object before the brakes could be applied.

In attempting to further the development of a lever system that allowed maneuverability similar to hand rims, several prototypes were built that combined a brake with the drive system. Typically, the roller clutch was used for forward propulsion, and an ancillary lever or handle mounted on the main lever could be used to select neutral and then engage a friction clutch which could be used as a dynamic brake or when fully applied to rotate the wheels in reverse. Hand grip levers proved ineffective. Rotation of the hand grip was more promising, but a second hinge point on the lever that utilized inward motion of the hands proved to be a satisfactory control means and promised to be useful for quadriplegics and others with limited use of the hands.

Further experiments showed that selecting neutral, that is releasing the roller clutch, was not easily achieved and the operator was never quite sure if the system was in forward, neutral or reverse until the stroke was engaged. Meanwhile the clutch system was greatly improved, required little force and motion to engage or disengage.

It was suggested by Brubaker to disconnect the roller clutch and use the friction clutch for all three functions, forward, reverse and braking. The current prototype (Figure 1) is based on this concept. When the levers are in the normal or rest position, the clutch is disengaged. Inboard movement of the levers engages the clutch (see Figure 2) allowing propulsion in either direction or use as a brake. The clutch currently in use is a brake drum using external brake shoes with non-asbestos brake lining. Propelling the wheelchair is similar to rowing or skulling a boat, with the lateral motion of the levers comparable to feathering the oars so that they do not drag on the return stroke. This requires some practice for effective use, but once achieved allows maneuverability that is comparable to hand rims.

Figure 1

The UVa lever drive system mounted on a special built wheelchair. The levers are shown in the stowed position, resting on the 16 inch main wheels.

Figure 2

The UVa lever drive system showing the inward motion of the lever that engages the clutch.

It is not apparent in the present prototype, that propulsion is more effective or more efficient than hand rims for skilled users or athletes, although this may be a matter of choosing the appropriate wheels, mechanical advantage, etc. What is apparent is that new design possibilities emerge. For example smaller drive wheels can be used which an fit under the seat, for a narrower and shorter wheelchair with no obstruction in transferring. Further, the hands and clothing are less likely to be soiled. Perhaps the most important application is the use by quadriplegics and others with limited hand function. Using ski pole hand grips and elastic bandages, several quadriplegics have tried the system with favorable results, particularly in the access to brakes. Stops at the forward travel of the levers prevent the levers from being pulled away from the user when applying brakes. Thus steering and speed control while descending hills is easily accomplished. The need for a suitable hand grip for quadriplegics is indicated for further trials. The system described has merit in that it is inherently a light weight low cost system with excellent maneuverability, requiring only inward motion of the levers to select all functions.

Further work on levers is indicated to exploit other possibilities. Seeliger[4] has developed a system that allows power to be applied during the pull and push part of the stroke. This is a smooth efficient drive system, but must be disconnected for reverse and close-quarter maneuvering with hand rims. Further possibilities with the UVa system include the return to the double action clutch system but using inboard motion to activate the brakes. The Matsunaga wheelchair uses wrapped spring clutches and connecting rods from the lever to the hub. Forward motion is achieved by forward motion of the lever from a neutral position while reverse is engaged with rearward motion of the lever from the neutral position. No doubt many other possibilities exist and should be tried. The potential use for levers in wheelchair drive is apparent and continued investigation will lead to one or more acceptable systems which will enhance the mobility of manual wheelchair users.

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ACKNOWLEDGEMENT

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COLLISION-AVOIDANCE SYSTEM FOR A MOTORIZED WHEELCHAIR

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Abstract

A computer-controlled collision-avoidance system for motorized wheelchairs using ultrasonic ranging systems and variable speed motors has been developed. This system is intended to improve the safety of persons in wheelchairs and to improve maneuverability in close quarters. The system automatically provides deceleration when an object is detected in the line of travel while still allowing for fast and precise control of the wheelchair. The software controlled system is readily adaptable to the user's needs and can be retrofit to existing wheelchairs.

Introduction

Some handicapped or disabled persons must use motorized wheelchairs for mobility. Because these people sometimes lack the fine muscle coordination necessary to operate wheelchair controls precisely, they sometimes collide with furniture, walls, or other people, causing damage to property and injury to others or themselves. A collision-avoidance system built for a motorized wheelchair can prevent these problems by slowing the wheelchair to a safe speed when it approaches an obstacle.

Materials and Methods

The collision-avoidance system is a modification of a control system created by a previous design group in the Electrical Engineering Department at West Virginia University. The original control system is based on an Intel 8085 microprocessor chip. The microprocessor reads the present position of the joystick, determines whether acceleration or deceleration is needed, and outputs the appropriate speed and direction signals to two proportionally controlled motor units on the drive wheels. This system provides 16 discrete motor speeds and 64 acceleration steps, with 4 steps of acceleration between each steady-state speed.

The collision-avoidance system was added to this existing control system. The block diagram of the combined system is shown in Figure 1. The collision-avoidance system consists of three commercially available ultrasonic ranging systems and appropriate software to control the ranging systems and determine when avoidance is needed. A software flowchart for the system is shown in Figure 2. The ranging systems are positioned around the joystick control so that one transducer faces forward and one faces each side. Object distance is determined from measurements of the time between a transmitted signal sent by the ultrasonic transducer and the first echo signal received. A support system comprised of digital and analog components is used to interface the ultrasonic ranging systems with the microprocessor.

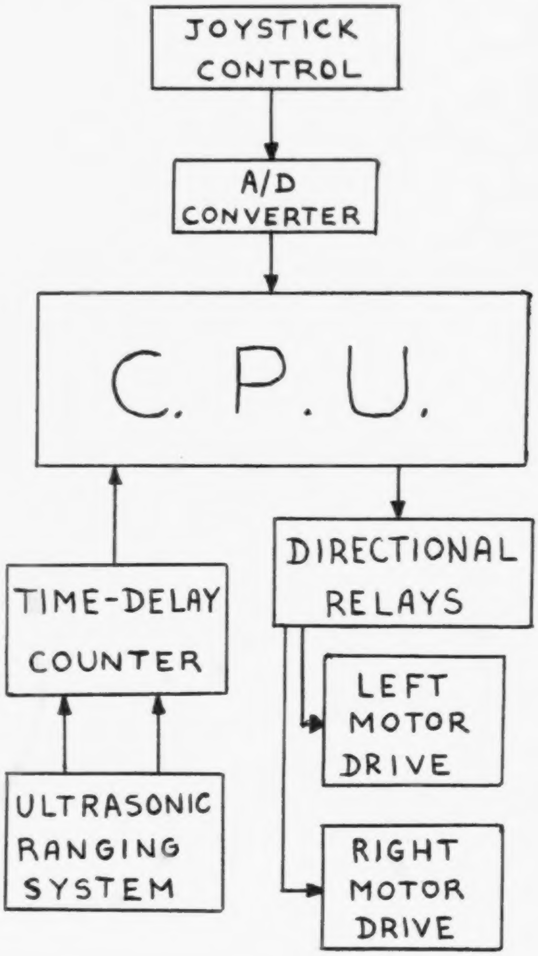

Figure 1 System Hardware Block Diagram

The control system allows the wheelchair to travel at full speed as long as there are no objects within 1 foot of either side and 6 feet in the direction of travel. When an obstacle is detected within these distances, the wheelchair is slowed to a safe speed. The proportional speed control of the motors allows smooth acceleration and deceleration.

Results

The collision-avoidance system is designed to gradually decelerate the wheelchair if an obstacle 6 to 30 inches from the floor and 6 feet away is detected while the chair is moving at maximum speed. The 6 foot limit provides enough distance to decelerate the wheelchair while maintaining a smooth ride for the user. At chair speeds less than the maximum, the wheelchair is allowed to get closer to the object before automatic deceleration begins. The slower speeds allow good maneuverability in close quarters. The wheelchair is allowed to come in contact with objects at a maximum speed of 1/4 foot per second. This speed allows the user to safely contact a desk or table. Maximum speed can be obtained 1 foot from an object not directly in the line of travel so that the wheelchair can travel at maximum speed down a narrow hallway or near a wall. When an obstacle is detected within 1 foot, the wheelchair immediately slows to 1/4 foot per second. Once the obstacle has been removed, the wheelchair automatically accelerates to the speed set by the joystick control.

The beam patterns transmitted by the ultrasonic transducers are not wide enough to cover the entire area around the front and sides of the wheelchair as shown in Figure 3. Additional transducers or a different type would correct this problem.

Conclusion

This collision-avoidance system provides fast and precise control of a wheelchair. Because the system is software controlled, it can easily be tailored to the needs of specific users. Simple modifications in the software can change the maximum speed at which the wheelchair is allowed to come in contact with an object, the number of transducers to be used for object detection, or the distance from an object at which the avoidance system begins decelerating the wheelchjair. Since this is a complete control system, it can be used on any motorized wheelchair with only minor modifications.

Acknowledgments

We would like to thank Dr. Robert L. McConnell, Associate Professor, and Ms. Donna Anderson, Graduate Research Assistant, Department of Electrical Engineering, West Virginia University, for their technical advice during this project.

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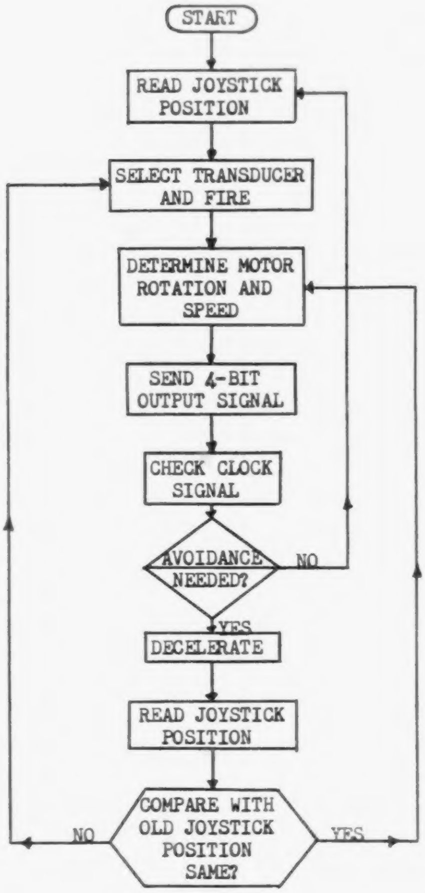

Figure 2 System Software Block Diagram

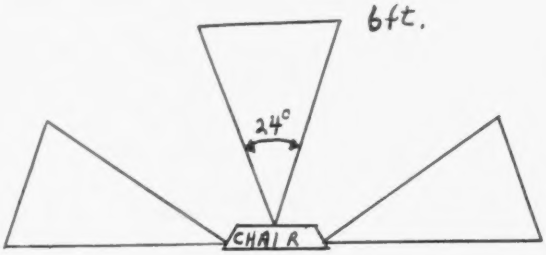

Figure 3 Beam Patterns

BLISSYMBOL TO JAPANESE CONVERSION SYSTEM

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ABSTRACT

A computerized communication system, to translate Blissymbols to Japanese text and speech, is presented. The system consists of a personal computer, and speech synthesizers to be attached to a Blissboard. The user can seem to speak Japanese merely by touching the Bliss board in the proposed communication aids.

INTRODUCTION

Recently, there have been active attempts to use Blissymbols easily with electronics and computer technology. The multi-language system, that translates the Blissymbols into English, Swedish, Spanish, Finnish, French, German, and Chinese , has already been developed. However, no proposal to make a Bliss-to-Japanese conversion system has ever been reported.

The authors are developing a 'Blissymbols to Japanese Conversion System'. The authors expect that this system will be added to the multi-language system as a Japanese version.

When Japanese people use the Blissymbols, there are many problems. One of the most serious problems is the fundamental difference in syntax between Western languages, e.g. English, and Japanese. That is, English is a 'Subject-Verb-Object (SVO)' type language, while Japanese is a 'SOV' type language.(Fig.1) Since the Blissymbols are based on English syntax, in order to translate the Blissymbols into Japanese, a syntax analysis is needed. For example, it is necessary to determine what a verb symbol is, which symbols constitute an object phrase, and so on.

Moreover, another difficulty is that a Japanese predicate phrase consists of a verb and auxiliary verb, and each word has many variations according to verb type, tense, voice, and sentence style such as interrogative or negative.(Fig.2) It is necessary to select the most suitable one among them. The authors have succeeded in solving these difficulties in this proposed system.

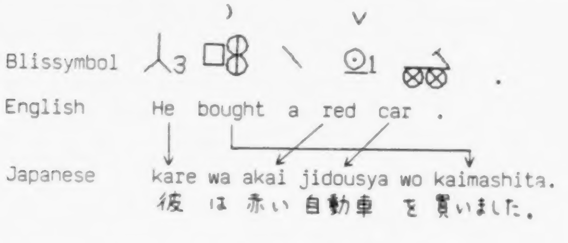
Blissymbol
English
He bought a red car .
Japanese
kare wa akai jidousya wo kaimashita.
彼 は 赤い 自動車 を 買いました。

Fig.1. English & Japanese Sentence Order

	English	Japanese
PRESENT	buy	kai masu 買います
FUTURE	will buy	kau desyou 買うでしょう
NEGATIVE	don't buy	kai masen 買いません

Fig.2. Japanese Verb Conjugation

GENERAL SYSTEM FEATURE

The authors' system consists of three commercially available devices, a personal computer, a digitizer, and speech synthesizers (Japanese/English) shown in Fig.3. The user can effectively speak Japanese using this simple system without any special attachment.

The user selects symbols on the Bliss board by the help of the digitizer. Selected symbols are converted into assigned codes to each symbol. They are analyzed in the Analysis Block by Bliss syntax and dictionary information. The Translation Block generates well-formed Japanese sentence according to Japanese syntax and the information stored in the Analysis Block. The Output Block converts Japanese sentence (ASCII) into codes that the Japanese synthesizer can accept, and adds intonation codes to them.

Editor/Filer manages Bliss texts and the dictionary. It updates symbol information and displays it, and appends new symbols to the dictionary.
The dictionary contains approximately 400 symbols. Each symbol is allocated a 256 byte space. The total stored information amounts to 100K bytes. The System manager supervises the entire system. (e.g. choosing mode(word or sentence), executing file-management program, selecting the way symbols are entered (from digitizer or file))

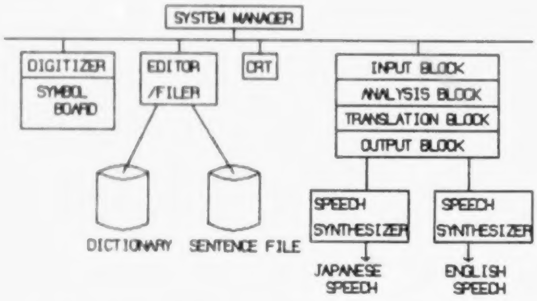

Fig.3. Bliss-to-Japanese Conversion System

ANALYSIS BLOCK

The Analysis Block analyzes the Bliss sentence according to Bliss syntax rules and dictionary information, and extracts the data, which are needed for translating to Japanese, such as part-of-speech, phrase, sentence style, tense, and voice.(Fig.4)
In addition to these processes, for translating more complicated Bliss sentence into Japanese, some sophisticate processes are required, such as an idiom analysis or a syntax meaning analysis. However, as the first step, these features have not yet been supported.

Word Analysis Block

This block analyzes contextual parts-of-speech in a Bliss sentence. In addition, it processes indicators, pronouns ($\lambda_3 \longrightarrow$ he), possessive cases ($\lambda_{3+} \rightarrow$ his), and so on.

Phrase Analysis Block

This block divides the Bliss sentence into several phrases with the parts-of-speech information. Five phrases can be accepted. They are :

- Noun phrase (NP)
- Adverb phrase (AdvP)
- Adjective phrase (AdjP)
- Preposition phrase (PrepP)
- Verb phrase (VP)

Subject Analysis Block

This block extracts the main data from the subject, and analyzes person and plural/singular data for the sentence.

Verb Analysis Block

This block analyzes tense, voice (active or passive), and verb type(transitive or intransitive verb or 'to be').

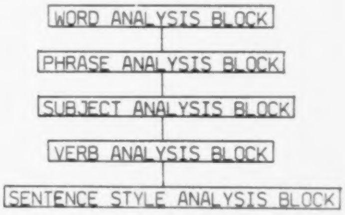

Fig.4. Analysis Block

TRANSLATION BLOCK

This block consists of three blocks, the subject conversion block, verb conversion block, and predicate conversion block.(Fig.5)

Verb Conversion Block

This block has the general verb conversion part and the BE conversion part. These are selected by verb type.
The BE conversion part conjugates Japanese "desu", which means 'to be' in English, according to a morphological rule. The general verb conversion part produces the correct inflectional ending of a Japanese verb, and adds the most suitable auxiliary verb.

Predicate Conversion Block

This block arranges the converted phrases in Japanese sentence orders. Three sentence sequences are provided, that is, SV, SVC, and SVO, as the orders.

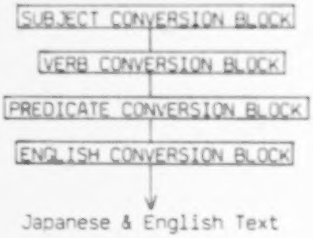

Fig.5. Translation Block

CONCLUSIONS

The proposed Bliss-to-Japanese system, in spite of its simple structure, allows users to easily compose many well-formed simple Japanese/English sentences. Furthermore , it is a bilingual system (English and Japanese), which handicapped Japanese people can communicate with English speaking persons by the aid of this system.
However, in order to apply it for educating handicapped Japanese, there are still some unsolved problems. One of the most serious problems is the difference in syntax between Blissymbols and Japanese. To solve this problem, it may be necessary to prepare a Japanese-oriented-Blissymbol, in which the orders are rearranged according to Japanese sentence. However, Japanese-oriented-Blissymbols will lose one of the most important features of using the Blissymbols, world-wide commonness of expression. Improvement in this system will lead to one solution for this problem.

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4.2

A MICROCOMPUTER ADAPTIVE CONTROLLER FOR ELECTRIC WHEELCHAIRS

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ABSTRACT

The construction and evaluation of a digital adaptive controller is described in this paper. This system includes test facilities which simulate actual wheelchair operating conditions.

INTRODUCTION

Wheelchairs experience a wide range of load parameter variation depending upon the mass of the wheelchair rider, the position of this mass in the wheelchair and the type of surface the wheelchair is traveling on. A velocity feedback controller improves the behavior of an electric wheelchair; however, the wheelchair's response can be significantly degraded by the time varying loads it encounters during its operation.

Traditionally, velocity feedback controllers are designed for some nominal load; hence, their undesirable response characteristics under changing load conditions. An adaptive controller was suggested to overcome the effects of large load parameter variations encountered by an electric wheelchair [1].

A variable structure adaptive controller was designed and simulated at the University of Virginia. This type of controller can provide an optimal response for a range of load conditions by changing its control coefficients as new loading situations develop. The purpose of this paper is to describe the system designed to implement and test the adaptive controller on an actual wheelchair motor under known load conditions.

BENCH TEST FACILITIES

Before attempting field tests of the adaptive controller, a bench test system was designed to provide maximum flexibility during the implementation process. This system allows us to create a simulated operating environment for the motor with known load parameters whose effects can be carefully monitored and analyzed.

There are four main components in the bench test system. They are a Magtrol Dynamometer, Magtrol Power Analyzer, Magtrol Dynamometer Readout and an armature controlled dc-motor. The interconnections between these devices are shown in Figure 1. With this assembled apparatus, a motor can be monitored for information such as speed, power consumption and armature voltage and current under prescribed loading conditions.

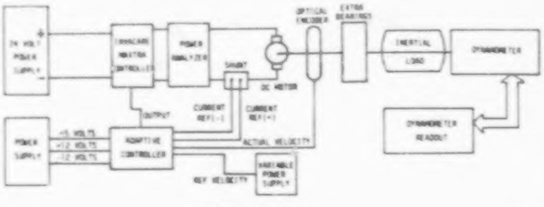

Figure 1

ADAPTIVE CONTROLLER HARDWARE

A Z80 microprocessor was chosen to implement the adaptive controller for several reasons. First, there is an abundant supply of Z80 support products available at a reasonable cost. Secondly, the software for the controller was written in FORTRAN on a Z80 based machine; therefore, the purchase of a high level language complier was unnecessary as long as a Z80 was used in our design.

Initially, a Z80 based system was conceived specifically for the adaptive controller, but the expansion of such a system was deemed too limited for future digital controller experiments. A commercially available Z80 system configured on a STD bus was purchased, because several manufacturers offer Z80 processor boards and Z80 peripheral boards with this type of standard bus.

To perform adaptive control, armature current, shaft velocity and desired velocity must be sampled by the system. This data acquisition task is handled by a custom designed board which uses an 8-channel, multiplexed, 8-bit analog-to-digital converter (ADC). Each input signal, moreover, is conditioned to meet the required analog amplitude range of the ADC.

Armature current is measured by a 25 ampere shunt connected in series with the armature of the motor. This shunt has an output range of 0-50 millivolts which is adjusted to represent -25 to 25 amperes through the ADC.

The actual velocity is measured by an optical encoder mounted on the motor's shaft. The encoder produces pulses at a frequency proportional to shaft velocity by directing infrared light off a disc encoder surface and sensing the presence of holes on the disc. These pulses are converted to a useable voltage representation of velocity with a frequency to voltage converter.

At this time, the desired velocity information is provided by a variable power supply because the adaptive controller operates independently of the directional input device selected. Eventually, a more sophisticated user input device will be tested in conjunction with the controller.

The power source for electric wheelchairs is usually a pulse-width-modulated (PWM) controller. In our test bench an Invacare Maxtra PWM controller was used for this purpose. The transfer characteristic of a PWM controller is non-linear, but as a first approximation it can be substituted by a linear transfer function for analysis. The transfer function of the Maxtra PWM controller was obtained from the measured input/output characteristics for the velocity range of the adaptive controller.

The command signal for the Maxtra PWM controller is usually generated by a joystick. In our system this signal must come from the adaptive controller through a digital-to-analog converter (DAC). Once the joystick input to the PWM was disconnected, the output signal from the adaptive controller was connected in its place.

TEST RESULTS

Before testing the adaptive capabilities of the controller, the performance of each of the four control equations was evaluated to verify their control coefficients. This was accomplished by observing step response data recorded for a measured inertial load with each control equation applied to this load separately. The test results yielded acceptable performance from each controller for inertial loads in their respective operating region.

Since the region of operation is mostly dependent upon the inertial load [1], the performance of a controller was evaluated for a range of damping loads while keeping the inertia constant. The performance criteria for this aspect of the controller was the ability to maintain a desired reference level despite load damping variations. Each controller showed excellent velocity control, and a comparison with the Invacare Ranger controller under the same conditions is shown in Figure 2.

The estimates of load inertia and load damping are the most critical calculations required for adaptive control. A recursive least squares approach is used to compute these estimates [1]. Although the load parameter estimates converged properly in the software simulations of the adaptive controller [2], they do not converge correctly in the bench test system.

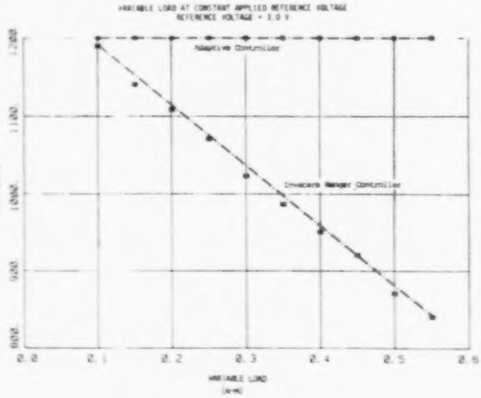

Figure 2

There are two major causes for this problem with the estimator which were verified by additional software simulations of the controller. The first difficulty arises because the estimator performance is directly related to the sampling rate of the system. Unfortunately, the sampling time selected for the original simulations is too fast for realistic implementation. Hardware and software modifications have been made, however, to permit sampling times which produce acceptable estimator results.

The second and more damaging problem is the measurement of armature current. A step input signal to this control system produces a large instantaneous current in the armature of the motor. In the present adaptive controller setup, current values are measured up to 25 amperes and then the measurement saturates at this value. The current values used in the software simulations range from 40-150 amperes during the initial phase of the step input waveform. If the current is limited to 25 amperes in simulation, the load parameter estimates do not converge as expected.

After the input to the controller has stabilized, current values remain well within the measurable 25 ampere range of the system, regardless of the load conditions. The estimate of the inertial load, however, is critical only during periods of acceleration. Therefore, a more accurate measurement of current during a reference input change is necessary for the recursive least squares algorithm to function correctly. Several possible solutions to this problem are presently being considered.

Once the estimator difficulties are overcome, the adaptive controller will be installed in an electric wheelchair. Extensive field tests will then be performed to prove the advantages of adaptive control over present control methodologies.

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This work was supported by NIHR Grant #G00-83-00072 at the University of Virginia Rehabilitation Engineering Center, P.O. Box 3368, University Station, Charlottesville, Virginia 22903. Phone (804) 977-6730.

4.3

MORSE CODE COMPUTER INPUT USING THE MOD KEYBOARD

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ABSTRACT

Morse code input techniques provide a useful, efficient data input technique for severely handicapped individuals. This paper describes a new version of the MOD Keyboard that provides the option of Morse code input.

INTRODUCTION

The MOD Keyboard system (as described in (1)) consists of a video display, a VIC-20 computer and a special cartridge which plugs into the VIC-20 computer's game slot. This cartridge converts the VIC-20 into a versatile aid for the severely handicapped that provides the following features:

1. A communication device which allows a variety of input devices and selection techniques. The screen of the VIC-20 computer is used to display a matrix of characters, words and phrases from which the user makes selections. Using one of several selection techniques, a handicapped user may build messages that may be sent to a voice synthesizer, printer, or read directly from the video display screen.

2. User alterable selection screens are available. The user may create unique screens for selection of individual characters, words or phrases. Once these pages are created, any one may be recalled and scanned for selection. All pages created by the user are stored in battery-backed CMOS RAM in the MOD Keyboard cartridge itself. Thus these pages are available whenever the MOD Keyboard is turned on.

3. Simple text editing is built-in. Without connection to another computer, the MOD Keyboard may be used as a simple text editor for writing letters or memos.

4. Computer keyboard emulation is an inexpensive addition to the MOD Keyboard system. Although keyboard emulation is built into the MOD Keyboard operating system, an adapter is required for each computer supported (ranging from a simple cable to an inexpensive adapter box (about $50)). Instead of entering keystrokes with the standard keyboard of the following computers: IBM PC, Apple II+, Apple IIe or the Commodore 64, the MOD Keyboard can be used as the input device. This feature allows a user access to the scores of software packages designed to run on any of the target computers. In each case the standard keyboard of the target system remains active while the MOD Keyboard may be used as an additional input device.

In summary, the MOD Keyboard provides a means for a wide variety of control devices and selection schemes. Its uses range from a simple communication aid to a full-blown programmable computer keyboard emulation device.

Morse Code Input

Presently, the MOD Keyboard is operated by scanning and selecting items from the screen. Although scanning is cognitively easier to learn than encoding, some severely handicapped people are able to learn Morse code as a means of character entry. For those who can learn to use it, Morse code character entry has been shown to offer greater mechanical and motoric efficiency over scanning techniques (2). Mechanical efficiency is inversely proportional to the time required to select an item and motoric efficiency is inversely proportional to the number of motor movements required to make a selection. In other words, it is both faster and requires less motor activity to use Morse code for character selection when compared to scanning techniques. To accomodate the segment of the handicapped population that can learn to use Morse code efficiently, a Morse code version of the MOD Keyboard has been developed. Although others (2), (3) have developed Morse code input methods for computer keyboard replacement, the MOD Keyboard offers great flexibility in terms of enhanced text entry speed via the word/phrase selection concept and the use of screen feedback of character codes as they are entered.

METHOD

The MOD Keyboard cartridge contains one 16K byte EPROM (to house the MOD Keyboard operating system), 8K bytes of battery-backed CMOS RAM and a parallel interface device (6522 VIA). No hardware changes were required to the cartridge to support Morse code data entry. A different program residing in the cartridge EPROM is all that is required to accomodate the special requirements of Morse code input. The MOD Keyboard arrangement has allowed the development of several versions of the MOD Keyboard by simply rewriting the MOD Keyboard operating system.

The original version of the MOD Keyboard is strictly a screen selection device. In the case of Morse code we are dealing with a serial coding selection scheme. Even though the screen matrix character selection strategies, originally used by the MOD Keyboard, are of course not applicable there are several benefits in using the video display of the MOD Keyboard system, these are:

a) The screen provides feedback of Morse code keying. This helps when a user is first learning Morse code and is later useful (along with audible feedback) in pacing Morse key data entry.

b) Built-in text editing features allows simple letter writing without an additional computer.

c) Two-character entry for selection of a word or phrase from lists presented on the MOD Keyboard Screen is possible.

d) A display device is required for the MOD Keyboard's built-in editor for editing the user selectable words and phrases.
The Morse code sequences used in the MOD Keyboard are shown in Table 1. Several special key sequences have been added to the standard set to allow the selection of the large variety of keys and key combinations available on many of today's computers. For instance the shift sequence (six dashes,------) puts the MOD Keyboard into upper case mode for the next character selection. Sending the shift sequence twice is equivalent to shift lock. A similar arrangement exists for the control key sequence. The repeat sequence (----**) is used to repeat the next entered character. PF1 and PF2 are two programmable function keys. When either one of these keys is selected, the MOD Keyboard displays a list of selectable functions. These range from Set Up, Key Editor, to any of the preprogrammed words or phrases that appear in a list for user selection. The selectable words are listed in two columns and have associated with them a single select character. If one of these select characters is entered, the word associated with that character is entered as though it had been directly entered character-by-character by the user.

Set Up

Once the Set Up mode is entered the user is presented with screen menus of options which may be altered to suit the configuration of his system. The main options included are:

1) Key response. How fast the Morse keyer accepts new entries (ie. the speed of "dit" and "dah" entry)

2) Transmit mode. Included here are several options for character storage and transmission to a computer or printer.

3) Output interfaces. This option enables the user to select the type of output device (ie. computer keyboard to be emulated or printer) that is to be connected.

Several other options are included to change some of the MOD Keyboard's operating features.

Key Editor

As mentioned above, the programmable function (PF) keys allow the selection of a word or phrase from a list displayed on the MOD Keyboard screen. Only the first 8 characters of the word are displayed on the PF screen. When a particular selection is made, as many as 254 characters may be entered by MOD Keyboard. The word displayed on the PF page may be either the first eight characters of the character sequence, or it may be a word (label) that stands for a sentence or operation that will be entered by the MOD Keyboard. This means that a selection labeled UP may be used to enter ESC D (where ESC is the escape character). In the Apple II computer this sequence is used to move the screen cursor up one line while editing Applesoft BASIC programs. In another example, the user may program a key to enter "Dear Sir or Madame:", even though the PF screen entry appears simply "Dear".

These programmable selection pages are edited using the MOD Keyboard's built-in key editor. As in the Set Up mode, the user is prompted for appropriate entries using a variety of menus. In this manner, a key may be programmed to generate a wide variety of responses. The key editor also may be used to redefine actual Morse code sequences required to produce a variety of either single or multiple character responses. Hence, frequently used words can each be given their own unique "Morse" code.

User Feedback

Morse code entry is accomplished by a two switch sequence, similar to that available with commercial Morse code keyers. One switch is used to send a "dah" and the other sends the "dit". Holding one or the other switch closed will send a stream of "dits" or "dahs". As with commercial keyers, there is audible feedback of the dot or dash as it is entered. The "dit" is a high pitched tone and the "dah" a low pitched one. A very helpful feature of the MOD Keyboard is the visual feedback provided on the MOD Keyboard display screen. The bottom line of the display screen is used as a status line. On this line the user can see the state of the Upper Shift or Control Shift keys as well as the "dits" and "dahs" as the current currently Morse code is being built up.

DISCUSSION

The MOD Keyboard is a highly versatile, inexpensive communication and keyboard emulation device. Adding a Morse code input version to the MOD Keyboard family (4) provides another option from which a user or therapist may select an optimal communication technique. Features such as: visual as well as audio feedback of Morse code entry, multiple character entry with only one or two

Morse code characters, and the keyboard emulation and word processing capabilities of the MOD Keyboard, offers both flexibility and speed of character entry for the severely handicapped user. The visible lists of the entries corresponding to the programmable function keys are a great aid to the user's memory. Although other communication aids (5) for the handicapped provide for abbreviated keystrokes to enter frequently used words, etc.; they seldom provide "pop up" lists of these abbreviations the way the MOD Keyboard does.

a	*-	n	-*	1	*----	SPACE	**--
b	-***	o	---	2	**---	BACKSPACE	----
c	-*-*	p	*--*	3	***--	NEW LINE	*-*-
d	-**	q	--*-	4	****-	RETURN	---*
e	*	r	*-*	5	*****		
f	**-*	s	***	6	-****	SHIFT	------
g	--*	t	-	7	--***	CONTROL	******
h	****	u	**-	8	---**	REPEAT	----**
i	**	v	***-	9	----*	ESCAPE	**----
j	*---	w	*--	0	-----		
k	-*-	x	-**-	:	---***	,	--**--
l	*-**	y	-*--	;	-*-*-*	.	*-*-*-
m	--	z	--**	-	-****-	/	****--
PF1	***---			PF2	***-*-		

Table 1. A list of some of the Morse code characters supported by the Morse code version of the MOD Keyboard. This list is not the full ASCII character set. The SHIFT key sequence may be used to obtain many of the common ASCII characters, eg. SHIFT 1 will give the exclamation point or the list in Table 2 may be used.

!	*-**-	&	-*-**	$	*-*-*
#	--*-*	*	***-*	%	---*-
+	*-***		*****-	=	-***-
@	*---*	"	*-*--		-*--*
(	**--*	)	**-*-		-*---

Table 2. Some additional codes for direct selection of particular ASCII characters. These codes may be used instead of using the SHIFT sequence along with the appropriate code in Table 1 to obtain upper shifted characters. Characters not listed here may be added to the MOD Keyboard's Morse code vocabulary at any time using the Key editor.

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EXPANDED AND REDUCED KEYBOARD ADAPTATION FOR THE MOD KEYBOARD

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ABSTRACT

For every handicapped computer user, there seems to be a different character selection means to which he or she is best suited. Some handicapped individuals, for instance, prefer specialized direct selection devices in the form of either expanded or reduced keyboards. This paper describes a version of the MOD Keyboard that supports a variety of these special keyboards as input devices.

INTRODUCTION

Computers have been seen as a way of providing expanded opportunities for severely handicapped individuals. Development of specialized devices and computer programs have enabled severely handicapped individuals to compete on a more even basis with non-handicapped individuals. Unfortunately most of these devices are either inadequate, prohibitively expensive or are only available in the laboratory. The MOD Keyboard system was developed to provide an inexpensive and flexible data entry device (1). Its main purpose is keyboard emulation (2). In other words, the MOD Keyboard is used to replace or operate in tandem with the keyboard of a variety of popular computers. Until now, the MOD Keyboard has offered scanning of its video display screen for selection of characters or words and phrases. Handicapped users able to manage a keyboard-like device prefer to use these devices rather than scanning methods for data entry. For these individuals, either expanded or reduced keyboards have been developed.

Expanded keyboards consist of a large number of switches, each with oversized keytops (2.5 cm. diameter) widely spaced over a much larger working area than a standard computer keyboard. These keyboards are useful for those with spastic disorders or who can only use their feet. Reduced keyboards look like miniature versions of a standard computer keyboard. Patients with low wrist or finger mobility (caused by multiple sclerosis or arthritis) find this type of arrangement very useful.

There have been a number of expanded and reduced keyboards developed for a variety of computers (3). The MOD Keyboard has now been adapted for these keyboards to provide a system that offers the following features:

1) Built-in local text-editing. This allows a user to perform simple text entry without requiring the attachment of another computer.

2) Screen display for user feedback. Quite often a keyboard user must be aware of the condition of the keyboard, ie. whether any of the various shift keys is in effect. With the MOD Keyboard system, a separate video display screen is used in the control of the keyboard. To reduce the amount of changing of view between the MOD Keyboard's display and that of the target computer, all data entered at the keyboard are displayed on both video displays.

3) User selectable and user alterable words and phrases. Much effort and time can be saved during text entry by providing user programmable keys. With this facility, one or two key presses may be used to enter several commonly used words or commands.

METHOD

The MOD Keyboard system consists of a VIC-20 computer, a video display screen, a MOD Keyboard cartridge, and a user's interface (special keyboard).

Hardware

The most important ingredient of the MOD Keyboard System is the cartridge itself. It contains the following (see Figure 1):

1) an EPROM (Erasable, programmable, Read-Only Memory of 16K bytes capacity),

2) battery-backed CMOS RAM (2K bytes), and

3) a Versatile Interface Adapter (VIA, 6522).

The VIA chip controls 20 signal lines that appear on an edge connector coming out of the back end of the MOD Keyboard cartridge. In this design, the user's keyboard attaches to this connector.

The keyboard itself consists of an array of switches arranged in rows and columns. With the the present design of the MOD Keyboard, the maximum number of switches that may be accommodated in a keyboard is one hundred, formed in an array of 10 rows by 10 columns. As with other versions of the MOD Keyboard (4), the only change required in the MOD Keyboard cartridge was the rewriting some of the software residing in the EPROM.

Keyboards

A variety of keyboards may be used with the MOD Keyboard. Currently the expanded and reduced matrix keyboards from TASH (Technical Aids and Systems for the Handicapped) are

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supported. (Other keyboards may be easily added as they are requested.) Although the keyboard's keys may be arranged in the standard QWERTY format, other formats are possible. One alternate arrangement is depicted in Figure 4. Here the keys are arranged such that the most often used keys are towards the center of the keyboard.

Programmability
Even though direct selection is an efficient method for character entry it is still a slow process for many handicapped individuals who may be using a toe, hand splint, mouth stick or headstick. To speed data entry, the MOD Keyboard has been designed to allow the user to enter a whole word or phrase by pressing only one or two keys.

There are 64 keys on the TASH keyboards; 47 keys are used for alphabetic, numeric, punctuation and other symbols, 10 are used for special characters (space, carriage return, escape, etc.), 3 keys are used as shift keys (SHIFT, CONTROL, CAPS LOCK) and 4 keys are function keys (F1 to F4). Any of these function keys may be used to select one of the following operations:

1) Set Up mode may be selected for changing a variety of parameters controlling the operation of the MOD Keyboard. The user may alter the switch delay for the keyboard (time between the physical switch closure and the acknowledged switch closure), the type of computer keyboard to be emulated, and several other parameters.

2) A Key Editor may be entered to change a key definition.

3) Predefined words or phrases may be selected from a list displayed on the screen.

Key Editor
With the Key Editor, a user may generate his own lists of words, phrases or special functions for later use. A word or phrase may be stored "under" a single key. Once a function key is pressed, the list of words stored "under" it is displayed on the screen. Each word has associated with it a keyboard character which is actually used to select the word. The words displayed on the screen are either the first eight characters of the message or simply a name representing the phrase which will be transmitted when its associated key is pressed. This labeling technique allows two keystroke selection of messages as long as 254 characters. There may be as many as 36 messages stored "under" each function key. Each function key, in turn, may be given a name that classifies the words "under" it. For instance, commands for spreadsheet programs may be stored under F2 with the name "VISICALC". While using the MOD Keyboard for normal text entry, a list of function keys and their names appears on the screen. This list provides a reminder of the types of words stored "under" the different function keys. In this fashion, a user may quickly access any word or command he has stored. As opposed to abbreviation expansion ([illegible]), this method of keyboard input acceleration requires very little memorization.

Several nested menus are used to prompt the user throughout the key editing process. Numbered options are presented in each menu. By entering the appropriate numbers, the user can quickly perform any of the following items.

1) Change the meaning of a single key to that of another. This is useful for trying different key position arrangements.

2) Change the meaning of a key so that its depression will perform local commands (such as shift or local editing commands).

3) Change the meaning of a single key to a string of characters. Some often used key sequences may be defined as a _single_ key depression.

4) Define a key for storage "under" a function key. Using this option, the user may set up a _two_ key sequence for selecting a character string (up to 254 characters long). Once the key is defined, the name appearing on its associated function screen is simply the first eight characters of the phrase stored by the user.

5) Define a key using a label "under" a function key. This is similar to 4) above except that an eight character label may be used to represent the string of characters.

Labeling of keys is particularly useful when emulating keyboards which have many more keys than the user's interface (ie. expanded or reduced keyboards). The user may define a label depicting the operation performed at the target keyboard. For example the function keys of the IBM PC keyboard may be labeled as F1 to F10 under one of the MOD Keyboard's function keys. The user may also define the key labels as the actual function performed by the depression of one the function keys (eg. "LIST" instead of F1, or "RUN" instead of F2, etc.).

DISCUSSION

Some handicapped people, given the choice and ability, would prefer direct character access over scanning methods. The version of the MOD Keyboard described in this paper provides that option. With its display, the MOD Keyboard provides visual feedback of the keyboard shift status. It allows the user to quickly and easily access a large number of words, phrases or target keyboard key combinations. Using its Key Editor, key positioning and spacing may be changed [illegible]. In general, this version of the MOD Keyboard expands the capabilities of the MOD Keyboard family by providing a very flexible direct access input means.

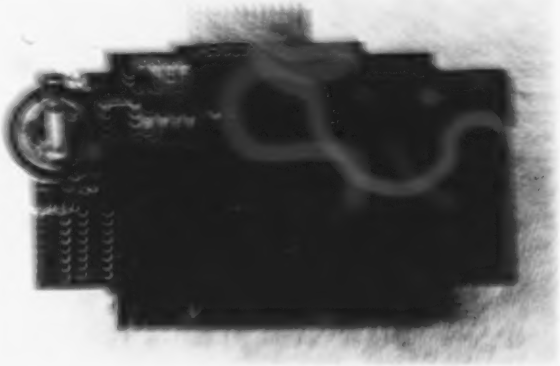

Figure 1. A photograph of the MOD Keyboard cartridge with its case removed. The wide edge connector at the bottom of the printed circuit board plugs into the VIC-20 game connector. The edge connector at the top plugs into the expanded or reduced keyboard via a cable.

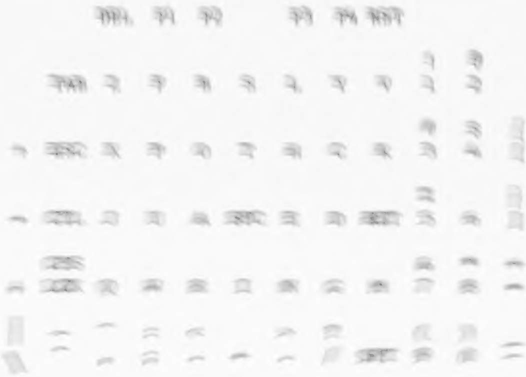

Figure 2. Typical layout for an expanded or reduced keyboard. Note the keys are layed out in accordance with the frequency of their use. Note the four function keys (F1 to F4) that are used to access the Set Up and Key Editor modes as well as word/phrase selections. (For this diagram the following abbreviations have been made: DEL-Delete, ESC-Escape, CTL-control, CPS LCK-Caps Lock, SPC-Space, RST-Reset, RET-Return and SFT-Shift).

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4.5

THE DESIGN OF AN ELECTRONIC DESK DICTIONARY FOR THE VISUALLY-IMPAIRED

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ABSTRACT

This paper describes a research and development project for visually and/or mobility-impaired individuals. The project involves development of a prototype electronic desk dictionary for such individuals. The initial design and an implementation schedule are described. Finally, completed work and future plans are discussed.

IDENTIFICATION AND SIGNIFICANCE OF THE PROBLEM

Currently, the severely visually-impaired must depend either on cumbersome Braille dictionaries, or the eyes of others, in order to enhance their respective vocabulary skills and abilities. This problem presents an opportunity to develop a device which would assist the visually-impaired in the utilization of a dictionary. The device would consist of:

(1) an audio input device;
(2) an appropriate microcomputer hardware and software system selected and designed for this particular problem; and
(3) an audio output device.

Such a device would facilitate much more rapid and less difficult access to a dictionary for the visually handicapped; this would greatly increase their individual learning potential throughout primary and secondary education. It would also be an invaluable tool for allowing the blind and near-blind to lead a more normal life as well as to make many career alternatives (such as technical writing, secretarial, and managerial positions) more realistic. The proposed device would be extremely useful also for most mobility-impaired individuals in ways similar to those that would make it worthwhile for the visually-impaired.

The proposed device would have many other significant consequences. It would be relatively easy to generalize the dictionary prototype to a foreign language or specialized technical dictionary (for instance, a medical, legal, or engineering dictionary). This would be very valuable to blind and mobility-impaired students while learning foreign languages or technical subjects. Due to their respective physical limitations, such handicapped populations could benefit enormously from devices, like the one proposed herein, which will make the educational process less cumbersome.

In addition, the development of such a device would find far-reaching applicability commercially in both the public and private sectors. All those involved in correspondence and report writing need to develop their respective vocabulary skills in order to be able to commit to writing what is actually meant. This requires a precise understanding of the meaning, and usage, of words as well as impeccable spelling. The type of device proposed herein would serve those purposes and lead to written documents of higher professional quality. The importance of this cannot be underestimated, particularly in a society very concerned with the decline in basic skills of its people. Furthermore, foreign language and specialized technical electronic dictionaries would have similar wide-spread commercial potential.

Initial Research Technical Objectives

The first goal would briefly be to investigate the feasibility of developing an electronic desk dictionary. This can best be done by developing a prototype since a number of questions must be answered in more detail than can be estimated without such a prototype. Specifically, answers are sought to the following questions:

(1) What is the most appropriate hardware system to support the software selected for the device?
(2) What is the most appropriate software to support such a device?
(3) How fast can a definition be retrieved in practice?
(4) How much will an actual system cost?
(5) How long will it take to transfer the information into computer-readable memory media?

Thus, the technical objectives will be as follows:

(1) to select an appropriate microcomputer hardware system, including determining necessary main and auxiliary storage sizes and speeds, input/output ports, and other hardware considerations;
(2) to select appropriate software for developing the system (such as a relational data base, sorting utility, or programming language);
(3) to obtain retrieval statistics about the prototype, which will be developed using hardware and software available to the author's company;
(4) to use the costs for the choices indicated by the selections made in (1) and (2) above, and other related costs to calculate the total cost of an actual system (because the central and auxiliary memory required may be extensive, the cost is difficult to estimate accurately without an actual working prototype); and
(5) to determine more accurate estimates of the data entry time required (this information will also be used in attaining objective (4) above, but is very important in its own right as it determines the time required for developing systems utilizing other written dictionaries).

Two other important technical objectives involve the determination of the most cost-effective (1) audio input device; and (2) audio output device. These objectives can be considered sub-goals of objective (1) above.

Utilizing the information gained through the achievement of the above objectives, further study into the feasibility, cost-effectiveness, and utility of the proposed system will be carried out. This will be the most important overall goal of the development project. In addition, further research into the feasibility of generalizations (foreign language and specialized technical dictionaries) will be carried out. This will be necessary for subsequent research, as described later.

Initial Research Work Plan

A stage-by-stage description of what will be accomplished and how it will be accomplished in the initial research will now be given. This will include a detailed discussion in each stage which addresses a specific technical objective, as outlined earlier in this paper. Each of the eight stages will now be discussed.

The first stage in the proposed work plan would include an organizational meeting, followed by a stage one work outline. This would include a literature and microcomputer product system review. Due to the rapid rate at which developments in the areas of audio input/output devices and microcomputer hardware/ software systems are being made, it is not practical to carry this out until work on the project is initiated. This stage would be scheduled for two weeks of work. The right to use the dictionary information of a major lexicographer would be secured.

This would be followed by stage two, in which the most current available audio input and output devices would be surveyed. The most cost-effective, high quality devices would be determined. This stage would be scheduled for a duration of three weeks.

The next stage would continue the hardware survey. In this third stage, about thirty general purpose microcomputer hardware systems would be investigated. The most appropriate five systems would be determined by a process of repeated elimination. This stage would be scheduled for a period of three weeks.

In the fourth stage the various applicable software products available for use as a system base in the design would be studied in detail. Four weeks would be needed: one week would be used to analyze dBASE II® (dBASE II is a registered trademark of Ashton-Tate); one week would be used to analyze SUPERSORT® (SUPERSORT is a registered trademark of Micropro, Inc.); one week would be used to analyze Microsoft's BASIC (a high-level language which has capabilities for both sequential and random disk access); and an additional week would be required to carry out a selection based on the combined analysis data of the preceding three weeks.

Then, stage five would last two weeks. A basic system implementation would be carried out. If more time were required, stages six and seven would be reduced in scope.

The following work would be included in stage six. Estimates for central and auxiliary memory size would be predicted at this point based on the preliminary system design and utilizing the data as keyed in up to this point in the project. Necessary input/output ports would be determined based on the audio input and output devices selected in stage two. Experiments with the prototype would be conducted to determine whether or not a sixteen-bit processor would be powerful enough to retrieve word definitions in a practical time. Two weeks are scheduled for this stage.

CURRENT PROGRESS AND FUTURE PLANS

This work represents a continuation of the research begun in (1).

A great deal of experimental work has been carried out to determine data entry time for input of a complete dictionary. Estimates for disk storage and access times, as well as microcomputer processor bit-size and memory requirements, for a system which will function at a reasonable rate of speed, are currently being determined.

Rights are currently being sought for use of dictionary information from several sources.

Audio input and output devices will be determined as the research progresses. State-of-the-art audio I/O equipment will be selected after the prototype is otherwise complete. Currently the system utilizes a Votrax® Model Number 100 (Type 'N Talk™) Speech Synthesizer (Votrax is a Division of Federal Screw Works) for audio output; no audio input device has been utilized as yet (the VET™ voice entry terminal is a possibility; VET is a trademark of the Scott Instruments Corporation). Since good audio I/O devices exist and will be improved considerably in the near future, the selection of such peripherals can safely be postponed while the remainder of the prototype is being developed. A good introduction to the technology of voice synthesis is contained in (2).

Additional dictionary query possibilities (besides spelling checking and definition determination) are being studied. Data structure alternatives to support such additional query options (such as synonyms or homonyms) are being investigated in order to arrive at an optimal implementation.

Pronunciation problems exist with the Votrax 100. If newer audio output devices do not solve these problems, accurate pronunciation may have to be achieved through phonetically spelling the dictionary data. This is obviously not a desired alternative.

Based on sample data entry, the dictionary information entry process is a slow one even with a highly skilled typist. Optical scanning has been considered for input but has numerous complications. Due to other priorities of the author's company, this project has not proceeded at the rate initially scheduled; however, an additional time commitment to the project is expected after the basic microcomputer processor and disk requirements are assessed. This work will be reported on at a later date.

AVAILABLE HARDWARE AND SOFTWARE

SUPERSORT (3), CONDOR (4), and BASIC (5) are being evaluated for implementation of this system. CONDOR is being assessed as an alternative to dBASE II; SUPERSORT is a leading sorting utility for microcomputers; CONDOR and dBASE II (6) are relational data bases (7). A Zenith Z-90 (8) system with 64K RAM, one inboard 100K hard-sectored single-sided, single-density disk drive and two outboard 320K soft-sectored double-sided, double-density disk drives is available as well as a Zenith Z-100 (9) system with 192K RAM and two inboard 320K soft-sectored, double-sided, double-density drives. CP/M (10) (a registered trademark of Digital Research, Inc.) and Z-DOS (11) (a trademark of Zenith Data Systems Corporation) are the operating systems currently being utilized. More powerful data base software, such as dBASE III™ (12) and METAFILE™ (13), are also being considered as well as totally integrated packages, such as Enable™ (14). [dBASE III, METAFILE, and Enable are trademarks of, respectively, Ashton-Tate, Sensor-based Systems, and The Software Group.] Besides data bases and sorting utilities, programming languages are also under consideration. In addition to BASIC, FORTRAN (15), COBOL (16), and Pascal (17) are being evaluated for possible implementation support software. Assembly language will be avoided unless timing problems cannot be overcome.

The Zenith Z-89 computer utilizes an 8-bit Z-80™ microprocessor (18) as its central processing unit as well as an auxiliary Z-80 processor for keyboard input processing. The Zenith Z-100 has both an 8-bit Intel 8085 microprocessor and an Intel 8088 (19) microprocessor (16-bit processor with an 8-bit data bus). [Z-80 is a trademark of Zilog, Inc.] Data entry has been carried out utilizing a special format in a word processor. This data will be transferred to different software systems during implementation.

ACKNOWLEDGEMENTS

The author wishes to thank George R. High for reviewing this paper and making a number of useful suggestions. The author also appreciates the enormous effort made by Shirley Hudkins in preparing this paper.

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ZERO PRESSURE KEYBOARD FOR APPLE 2E

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ABSTRACT - This paper describes a zero-pressure, computer keyboard developed at the Wesbrook Children's Technology Center to allow disabled individuals with weak muscular control independent access to microcomputers. The keyboard is designed to operate in parallel with the Apple 2e microcomputer keyboard while maintaining complete software transparency.

I. INTRODUCTION

Conventional microcomputer keyboards are unsuitable for many disabled persons with weak muscle disorders such as Spinal Muscular Dystrophy. For these individuals even the slightest finger pressure is very difficult or impossible. As a result, for the last year, the Wesbrook Children's Technology Center has been involved in the design and development of a zero-pressure, microcomputer keyboard to allow young children with such disorders easier access to microcomputer systems. The unit is currently available for Apple 2e users under the name KEASYBOARD and can be installed in less than five minutes.

Key selection is accomplished by inserting the finger through one of 65, 17/32" diameter holes and interrupting a pair of row and column light beams. Since no contact switches are involved, the user requires no finger pressure to access a key. Apart from the standard keyboard, an auxillary multiple character keyboard mode can be selected for sending several characters or words with a single key access. Customized auxillary keyboards are available for individuals who frequently enter the same words and phrases.

II. DESIGN

During the conceptual design phase of the project, apart from the zero-pressure feature, several objectives were considered important. The unit had to be inexpensive, easily adapted to existing, as well as future microcomputers, rugged, simply constructed and easy to install. Infrared light emitting diode transmitters and receivers were selected instead of capacitance type switches primarily because of lower cost. The entire microcomputer controlled circuitry is housed on a single 7 1/2" x 12" printed circuit board which is enclosed in a light weight plastic cabinet.

PHOTO 1: Zero-pressure computer keyboard for disabled individuals with weak muscle disorders.

The zero-pressure keyboard consists of a microcomputer controlled infrared scanner. There are 18 pairs of infrared light emitting diode (LED) transmitters and receivers mounted around the perimeter of the circuit board; with 5 along the Y direction and 13 along the X direction (as shown in figure 2) resulting in a total of 65 possible key points. These light beams are checked sequentually by the microprocessor for any interruptions. When a finger, head stick etc. interrupts a particular X-Y light beam pair, the key defined at those coordinates is selected. The hardware sequences each of the 18 light beams starting with pair Eo and Qo. In order to conserve power, only one beam is energized at a time. When any of the X beams are obstructed, the appropriate X binary code is then stored. The scan is then completed by scanning the Y beams in a similar manner and constructing the full 6-bit (X,Y) code corresponding to the selected key.

When the (X,Y) code has been constructed, it must be used to activate the appropriate key on the host computer, transparent to any software that may be running at the time. In other words, the host computer should not be able to tell whether the characters are coming from the KEASYBOARD or the regular host keyboard. In the case of the Apple 2e, this requires the KEASYBOARD output to appear like a matrix of 63 contact switches. The actual implementation consists of a bank of 64 semiconductor switches tied to the Apple 2e 10x8 line matrix. Each key is activated when a pair of matrix lines are tied together. An output code is fetched by the microprocessor within the unit through a table look-up procedure using the (X,Y) code obtained above. This 8-bit code is then sent to a decoder which is responcible for turning on the appropriate matrix switch. Since different manufactures may use different switch matrix configurations, different table codes and output cable configurations will be required, however, the principle still remains the same. Still other manufactures may output parallel or serial ASCII codes which would require some scheme of interweaving ASCII codes from both keyboards or simply disabling the regular keyboard. At present we have decided to restrict the KEASYBOARD to switch matrix interfaces since these microcomputers are more widely used today.

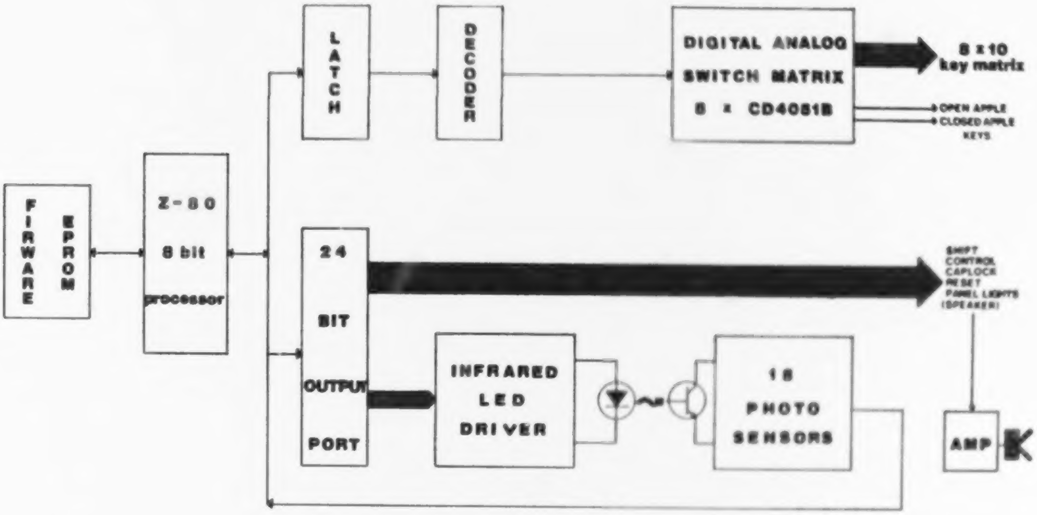
LATCH
DECODER
DIGITAL ANALOG SWITCH MATRIX
8 x CD4051B
8 x 10 key matrix
OPEN APPLE
CLOSED APPLE
KEYS
FIRMWARE
EPROM
Z-80
8 bit
processor
24
BIT
OUTPUT
PORT
SHIFT
CONTROL
CAPLOCK
RESET
PANEL LIGHTS
(SPEAKER)
INFRARED
LED
DRIVER
18
PHOTO
SENSORS
AMP

FIGURE 1: Block diagram of microprocessor controlled zero-pressure keyboard. Key selection is accomplished by interrupting infrared light beams which in turn activates one of 64 digital switches. These switches appear like the regular keyboard switch matrix to the microcomputer.

All the photodetectors are wired-OR together and read as a one bit status port. A twenty-four bit output port is used to strobe the 18 LEDs, control the two panel lights, activate the SHIFT, CONTROL and CAPLOCK lines, as well as generating an audio key "click" signal to the KEASYBOARD speaker. An additional, 8-bit output latch is provided for activating the required matrix switch. Six bits are sent to the matrix decoder while the remaining two bits control the OPEN-APPLE and CLOSED-APPLE keys directly. All the above control signals are assigned in firmware and can be altered for other microcomputer systems.

A powerful customized keyboard mode is available for individuals who frequently enter the same set of command words or phrases. Firmware tables can be set up with specific words and phrases during manufacturing to enable the individual to output word/phrases with only two key accesses(one key to select the auxillary keyboard mode and a second key to select the phrase).

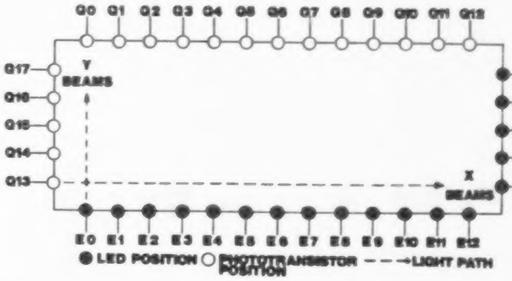
Q0 Q1 Q2 Q3 Q4 Q5 Q6 Q7 Q8 Q9 Q10 Q11 Q12
Q17
Q16
Q15
Q14
Q13
Y BEAMS
X BEAMS
E17
E16
E15
E14
E13
E0 E1 E2 E3 E4 E5 E6 E7 E8 E9 E10 E11 E12
● LED POSITION ○ PHOTOTRANSISTOR POSITION - - -→ LIGHT PATH

FIGURE 2: Infrared LED and phototransistor pair array giving 65 key selections on KEASYBOARD.

III. CONCLUSION

Initially five units have been constructed for preliminary evaluation on Apple 2e systems. Future work will involve development of personality modules to allow interfacing to Apple 2+, Apple 2c, IBM PC, Commodore 64 and Vic 20 microcomputers. The KEASYBOARD is currently being manufactured by:

Parallel Systems
P.O. Box 58435, Sta. L
Vancouver, B.C., Canada V6P6K2
(604) 261-4106

IV. ACKNOWLEDGEMENTS

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The Wesbrook Society (The Wesbrook Children's Technology Center)

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A MYOELECTRIC B/E PROSTHETIC SYSTEM FOR YOUNG CHILDREN

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Abstract

This paper describes a system of modular myoelectric control units designed for use with Systemteknik and Steeper electric hands. The development began with a myoelectric control designed for infants using the Systemteknik electric hand but was quickly expanded to incorporate myoelectric control modules designed for use with Steeper electric hands for young children. Design objectives include; improved cosmesis, reduced weight, ease of fitting, choice of two or three state control, and efficient use of battery power by the electric control module and the electric hand.

The resulting designs meet these objectives. The modular forearm segments are injection moulded of caucasian colored polypropylene. The built-in battery types have injection molded battery cases of fluoro plastics to withstand high temperatures and these types are rechargeable in 1/2 hour or less.

Background

From the experiences of Rolf Sorbye of Orebro Sweden, it became evident in the early 1980's that it was possible to fit very young children with myoelectrically controlled prostheses using existing adult controls and hands(1). Sorbye demonstrated that these young children could effectively control their prostheses at the young age of 18 months and that in later years their use was very spontaneous and natural compared to children fitted at older ages.

Introduction

In 1982, the Bio-Engineering Institute's Prosthetics Research Centre started a similar program of fitting young infants using adult sized myocontrol units from Otto Bock and UNB, and the Systemteknik hand developed in Sweden for Rolf Sorbye's patients. Similarily, we are seeing the same spontaneous and natural use of their myoelectric prostheses at these young ages(2). From our experience it was evident that existing controls used on these infants and young children were not appropriate in size or function as seen in figure A.

Modules for short and long B/E child amputees have been developed by the Institute. These control modules were designed with the following objectives;

a) improved cosmesis,
b) reduced weight,
c) choice of two state (two muscle) or three state (one muscle),
d) choice of built-in or external battery (fast charge option for built-in battery),
e) battery saver feature in both opening and closing.
f) ease of fitting (small electrode disks in the socket with electronic circuitry located below forearm laminating ring).

Materials and Methods

In conjunction with prototype evaluation of these new designs, individual design features were integrated with existing controls of other children and adult patients to evaluate their usefulness in the field. These features include battery saver and battery fast charge options(3).

All U.N.B. controls manufactured today have a built-in battery saver feature which decreases battery consumption by 1/2 when used by an active child. This feature is further expanded upon in our infant design model to save power on open as well as close.

During the past year a fast charge option for built-in batteries has been evaluated in extensive lab test and on patients already using UNB adult myocontrol units. This feature allows the patient to fast charge the battery in 30 minutes or less during the day while normal slow charge is used at night. For an added safety feature, the battery housing cases of these units are injection moulded of fluoro plastic to resist flame and high temperature.

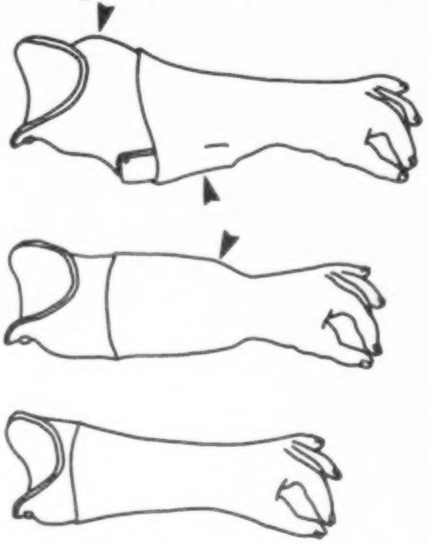

Figure A

(top) Otto Bock electrodes and externally mounted battery on Systemteknik hand,
(center) U.N.B. built-in battery on same hand,
(bottom) The new U.N.B. system for infants with Systemteknik hand.

Thick film hybrid technology is used on all the electronic modules to keep the physical size to a minimum and enhance the reliability. An example of an unencapsulated module is shown in figure B.

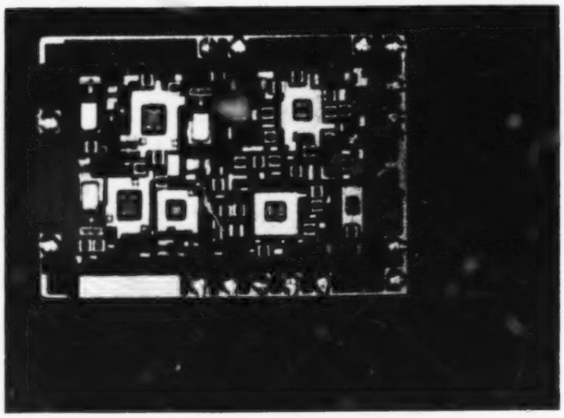

Figure B

All these electronic circuits are housed in an injection moulded, caucausian colored, polypropylene case that also makes up a portion of the forearm of the prosthesis. A laminating ring is incorporated at this level for the prosthetist to attach the socket.

All polypropylene cases attach directly to the different Systemteknik and Steeper electric hands as illustrated in figure C.

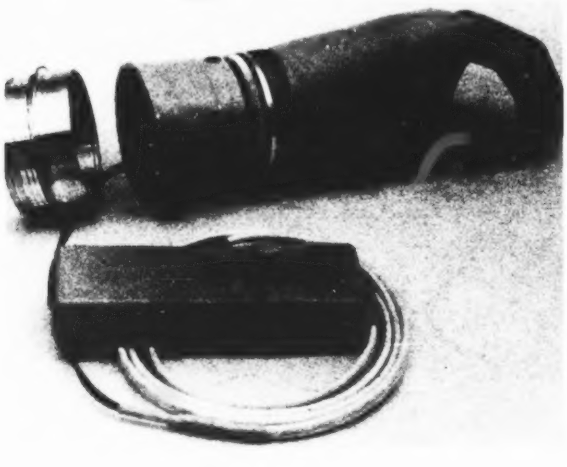

Figure C

At present we are designing two electronic modules to go with eight packaging systems for various levels of young B/E amputees. These packaging systems are listed in table on the next page.

Electric Hand	Three State Control	Two State Control
Systemteknik	1. Built-in battery	5. Built-in battery
	2. External battery	7. External battery
Steeper	3. External battery	6. External battery
	4. Built-in battery	8. Built-in battery

Conclusion

We feel that our modular design systems will allow numerous young patients to be fitted with cosmetically acceptable and electronically compatible myoelectric control units. The electronic modules can be transplanted from one modular case size to another to follow the child's growth through a series of larger hands.

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EVALUATION OF CABLE VS. HYDRAULIC TRANSMISSION OF FORCES FOR BODY-POWERED ARM PROSTHESES

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ABSTRACT

Different combinations of steel and plastic cable and housing were tested along with hydraulic master/slave cylinders to determine which are most efficient in the transmission of forces by arm amputees from the shoulder harness to the upper-limb prosthesis.

INTRODUCTION

This work is part of a larger project to assess the feasibility of improvement in body-powered upper-limb prostheses. The first step was a survey of arm amputees and professionals to verify needs and priorities (2). The work presented here is the second step.

IMPORTANCE

The survey mentioned above indicated that function is clearly the most important feature which arm amputees want and expect from prostheses. Uncomfortable harness and poor appearance are the most negative features of prostheses as rated by arm amputees.

The force transmission system is a key factor in the function, comfort, and appearance of arm prostheses. Obviously, the transmission of forces from the shoulder harness to the terminal device or elbow is critical to the functional operation. It is also very important to comfort because the harness takes high forces in the axilla and elsewhere. If the force transmission system were more efficient, the forces on the harness would be decreased and comfort increased. The force transmission is also important to appearance because amputees want all the hardware buried in the prosthesis, and it would be difficult to bury the present steel cable and housing without reducing the efficiency and thereby increasing the forces necessary for operation.

The work presented here was undertaken to evaluate whether the efficiency of the cable control system can be improved and if hydraulics offer a feasible alternative.

WORK OF OTHERS

Sammons (3) reported a 15% increase in efficiency in the steel cable and housing system by use of the Teflon liner. Lewis (4) and Goller (5) experimented with a hydraulic, body-powered system and shoulder shells vice harness, but this work was not continued. Heather (6) and Smith (7) developed an all-Nylon, compact, hydraulic system for operation of a "Helping Hand" orthosis, but this work was not continued either. Alderson (8) reported that IBM developed a hydraulic system operated by the foot, but it was considered cumbersome and was abandoned.

The work by Lewis/Goller and Heather/Smith showed some promise, and there is no clear evidence why these efforts were dropped. The author and others feel that the time is ripe to re-evaluate the use of hydraulics in light of current needs and technology.

TESTING

As shown in Fig. 1, a test was set up to measure the force of the following force transmission systems:

- Steel cable (Hosmer C-100) and steel housing (Hosmer CH-100)
- Steel cable and steel housng with Teflon liner (Hosmer CH-100HD and CHL-100)
- Steel cable and plastic housing (Parflex polypropylene)
- Plastic cable (Ledina 66 Tennis String) and steel housing
- Plastic cable and steel housing with Teflon liner
- Plastic cable and plastic housing
- Hydraulic master and slave cylinders using Clippard Minimatic components

A standard weight of 23.1 lbs. (10.5 kg.) was used to simulate the force required for terminal device operation. The weight was lifted measuring each of the force transmission systems on a straight line and then over a 2.25 in. (5.7 cm.) radius at bends of 90°, 180°, 270°, and 360° to simulate operation with various cable routings. Force to pull was measured with a dynamometer, and efficiencies were calculated.

RESULTS

The efficiencies of the various force systems are shown in Fig. 2. In every case, the plastic cable was equal or higher than the steel cable, and either cable in steel, unlined housing was the lowest. The best cable combination was the plastic cable with the steel housing and Teflon liner.

The hydraulic system has a built-in frictional loss in the piston seals in the cylinders, which was 10% in the case of the standard components chosen for test, However, the system efficiency remained at 90% for any routing of the hydraulic line.

CONCLUSIONS

The plastic cable riding in the Teflon-lined steel housing offers the highest efficiency of cable systems tested, being 90% at 180° and 81%

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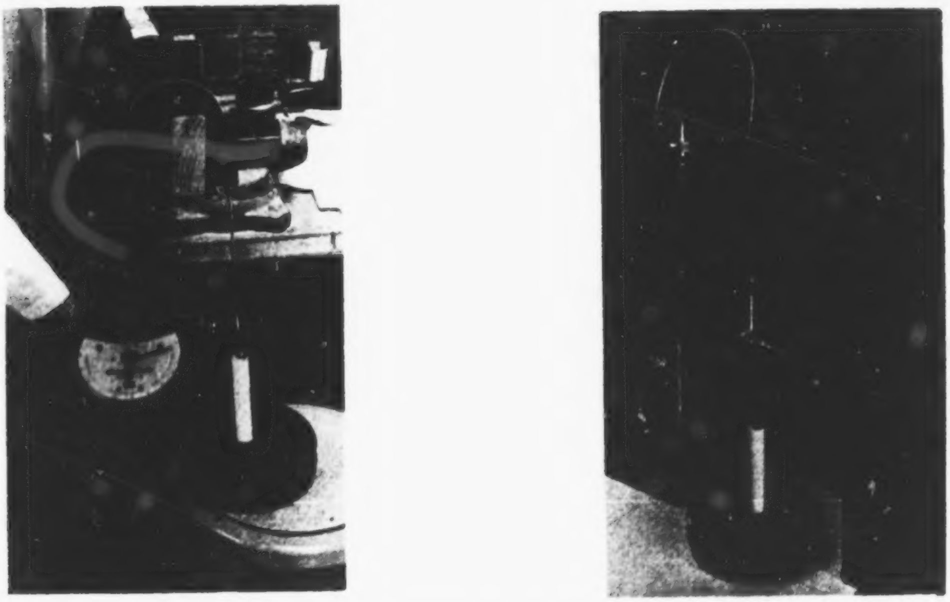

Figure 1. Test setup showing a cable system on the left and the hydraulic system on the right. Dynamometer used on all tests is shown on the left.

at 360° of cable routing. The durability of this combination remains untested.

The hydraulic system offers a constant 90% efficiency over any routing of the hydraulic line. Therefore, for situations where a cable must be routed through greater than 180°, the hydraulic system is superior in efficiency and would require lower forces by the amputee on the shoulder harness. Also the durability of a hydraulic system should be good. Smith tested his system successfully in a cycling machine for a period equivalent to 26 years of amputee use.

The hydraulic system appears to have a good potential for use as a force transmission system for body-powered upper-limb prostheses. As well as low frictional losses and high durability, it offers good sensory feedback and possibilities for mechanical advantage in force or excursion by using different sized piston actuators.

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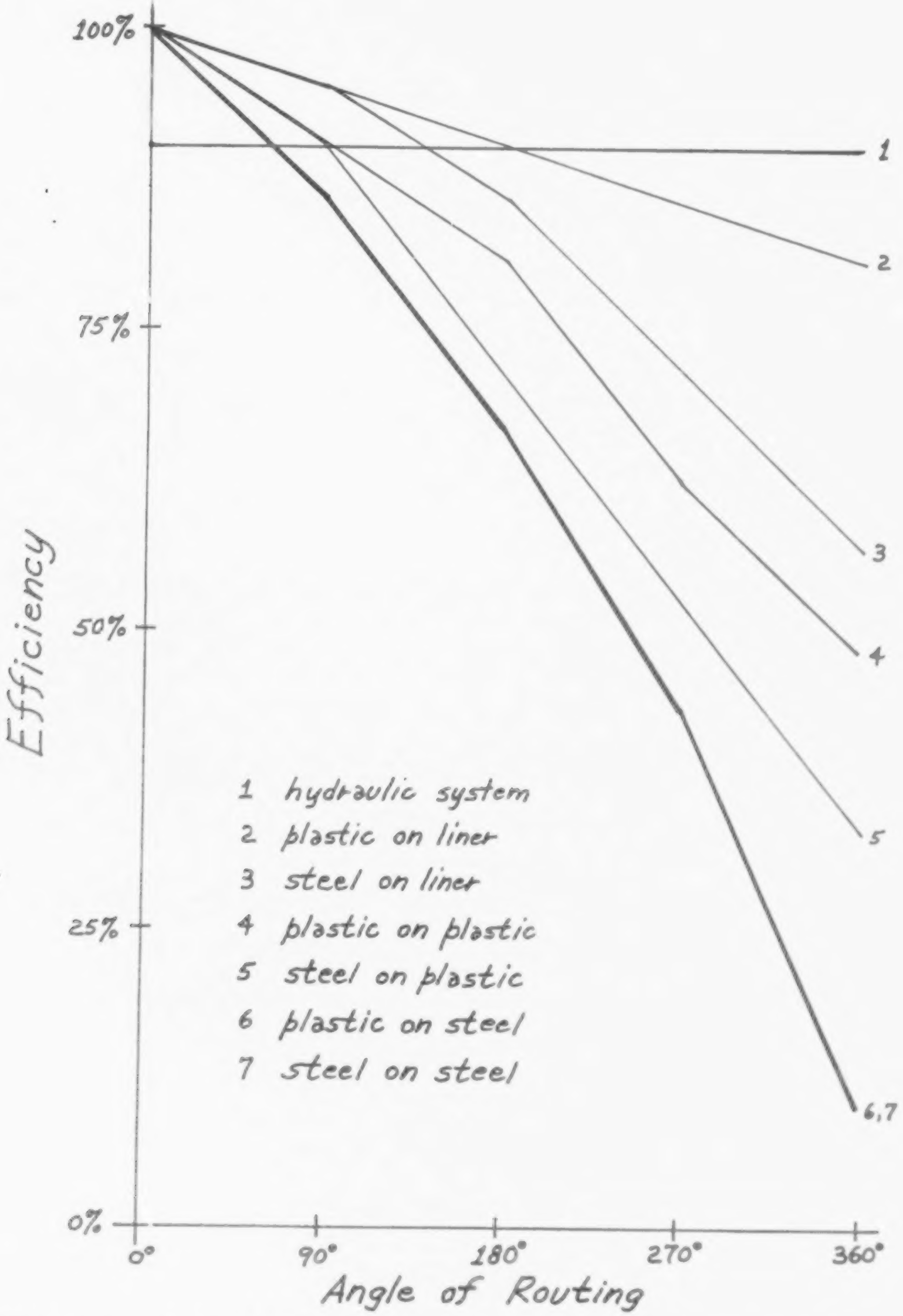

Figure 2. Results of tests showing efficiencies of force transmission systems at various angles of cable routing.

5.3

AN APPLICATION OF AESTHETICS IN THE USE OF A ROBOTIC ARM

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ABSTRACT

An assessment of attitudinal changes of selected quadriplegics in response to the presentation of two orientation programs on the use of a robotic arm was the focus of this project.

The whole study can be conceived of as a series of investigations of the hypothesis that aesthetic movement of a robotic arm can affect an individual's attitude toward the acceptance and use of robots. As a piece of exploratory research, this case study does not reveal conclusive evidence to support the hypothesis.

BACKGROUND

The choreography of synchronized robotic movement will be part of this present study which investigates user acceptance of the robotic arm in a rehabilitative setting. This case study features dance choreography for a robotic arm as part of an experimental treatment program in the attitude assessment study. Choreography, as demonstrated through aesthetic sequences of the robotic arm may enhance creative discovery in a non-utilitarian sense.

As part of this project I have attempted to present aesthetic movement of a robotic arm, in the form of choreographed sequences of robotic movement synchronized with music. The mechanical features of the staccato action of the robotic arm are contrasted with what I have defined as 'aesthetic' maneuvers. The aesthetic movements feature a more sustained effort in the actions, smoother transitions from point to point, curved lines replacing many of the straight and sharp angular motions, and a varied sequence in the timing of movement phrases to break up the constant speed characteristic of the practical patterns of movement. The aesthetic maneuvers explore the related movement elements of the action quality, flow, shape, and timing in various movement phrases.

In this project, a combined effort works to integrate the sounds of music, the forms of sculpture, and the motions of dance. The idea of choreography of synchronized robotic movement has been developed as part of the experimental aesthetic orientation session and demonstrates the aesthetic dimension of the robotic manipulator aid. Choreography, the art of making dances, uses dance as a series of rhythmic motions in time and space to express ideas through movement. The connotation of dance and choreography suggest direct human involvement or participation in the movement sequences; hence, the very idea of a robot dancing may seem self-contradictory. The human element is involved in the programming of movement sequences for the robot; however, a 'dancing' robot and robotic 'choreography' may be used only as metaphor. The process of choreographing for a robotic arm combines a logical approach with a sensuous approach in a blend of artistic-scientific creativity.

METHODS AND PROCEDURES

Seven quadriplegics, who may eventually make use of a robotic arm, were the subjects for this study. A coin toss divided the subjects randomly into two groups: Group A and Group B. All patients received a standard orientation program and one group also received a supplementary aesthetic orientation.

The Supplementary Aesthetic Treatment Session

The aesthetic orientation focused on a general introduction to the design of robots as part of the technological revolution in our society. This introductory session included individualized instruction, a slide presentation, and video-taped sequences of synchronized robotic movement. This program has been developed to reduce robot anxiety and increase robot acceptance by presenting aesthetic features associated with robotic movement.

The lecture provided a brief explanation of how robot technology has been developed and the robotic arm and was introduced during this portion of the presentation. The presentation also illustrated:

1) the evolution of robots from fiction to fact

2) the robot as part of advanced technology and an object of art

3) selected works of modern sculpture to illustrate the changes in materials and structures of art objects

4) principles of the Bauhaus School union of art and technology were emphasized through the work of: Kandinsky, Schlemmer, Moholy-Nagy

5) products of industrial design to illustrate the aesthetic design of utiliterian products

6) work of kinetic arts will be related to the 'machine aesthetic' as an outgrowth of the Bauhaus

7) the growth of the Bauhaus influence in the area of industrial design and architecture

8) the concerns of industrial designers in the aesthetic design of ordinary products will be compared to the movement of robots - video-taped segments of synchronized robotic movement choreograped for the robotic arm will be presented. The choreographed pieces included:

"Waves" - choreography by Margo K. Apostolos, music by Sandra Cotton

"Stardance" - choreography by Margo K. Apostolos, music by Sandra Cotton

"Freeflight" - choreography by Margo K. Apostolos, music by Gwendolyn Watson

COMMENTARY ON THE RESULTS

The investigation of the results presents the differences between two groups on each of the following variables:

1) familiarity with computers

2) use of computers

3) positive feelings toward computers

4) improvement in feelings toward robots

5) decrease in robot anxiety

6) increase in robot acceptance

Does the aesthetic movement of a robotic arm affect an individual's attitude toward the acceptance and use of robots? While the differences between the two orientation programs were not significant for all six variables, the scores are moving in the right direction. The aesthetic orientation appears to be more effective on the computer related variables.

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A MECHANISM OF ELECTRICAL POWERED SHOULDER ARM WITH A DOUBLE LINKAGE

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ABSTRACT

This paper reports a new type shoulder mechanism which can be minimized the torque of a driven motor to the arm's weight and the loads by using a double linkage mechanism.
Especially, for the physically handicapped person with both arms cut off, the total arm prosthesis is required as follows;

1) We have never developed a simple shoulder mechanism. If the function of arm increases, it will become heavy and large. And then, it will become difficult to model on natural arms for the external appearance.

2) This shoulder prosthesis has independently driven motors. It is very difficult to control freely for the handicapped person.

We have developed a new type shoulder prosthesis in consideration of these problems.

1) A double linkage mechanism to the shoulder joint is very simple, smart and functional.

2) This control system with one-tip micro-computer and new LSI and power IC are small. Consequently, this prosthesis can be controlled with only 6 switches for a new model control method.

SHOULDER MECHANISM

The actuator must be taken as large as the action of moment,³ as the multi-joint of robot, manipulator and total arm prosthesis are attended with an inevitable fault which has large action of the moment of inertia to the first joint.

Figure 1 shows the principal mechanism of a new shoulder prosthesis. The edge of ball screw is fixed to the socket by the coupler of swivel block, and the nut (D) is held to the connecting rod. When the movement of figure 1 is flexion and extension, the movements of abduction and adduction can be gained by the change of position at 90 degrees of the same mechanism of the swivel block in common (A).

The location of the shoulder with respect to the origin is defined by the motion of the elbow in space as shown in figure 2.

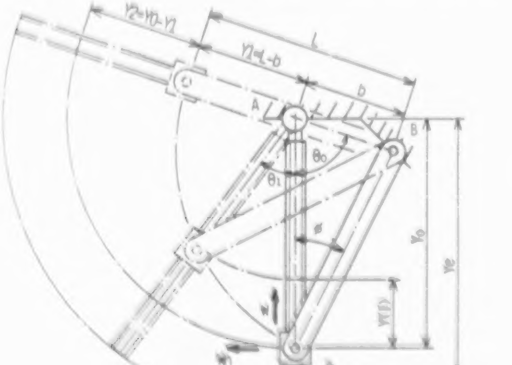

Figure 1 Principal mechanism

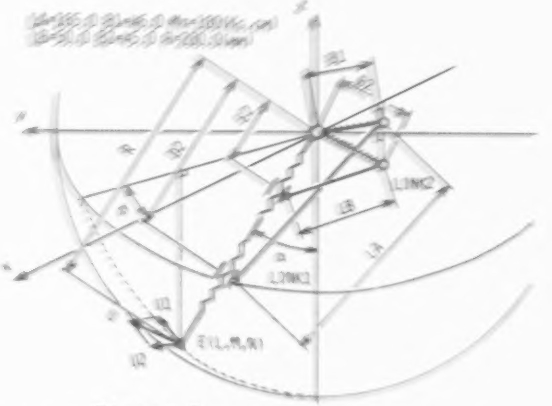

Figure 2

The components of the elbow position can be expressed as;

$$\begin{pmatrix} L \\ M \\ N \end{pmatrix} = \begin{bmatrix} R \cdot SIN\,\alpha \cdot COS\,\phi \\ R \cdot SIN\,\alpha \cdot SIN\,\phi \\ -R \cdot COS\,\alpha \end{bmatrix} \quad \ldots\ldots\ldots(1)$$

A position angle of α moves the upper arm to flexion, and position angle of ϕ raises the arm to abduction.
Where,

R = Length of upper arm

And moment of the elbow is given as follows;

$$U1 = D2/R \cdot M_T \cdot TAN(ACOS((LA^2+D2^2-B1^2)/(2 \cdot LA \cdot D2)))$$

$$U2 = E2/R \cdot M_T \cdot TAN(ACOS((LB^2+E2^2-B2^2)/(2 \cdot LB \cdot E2)))$$

Where, ……..(2)

U_1 : Moment of the shoulder flexion
U_2 : Moment of the shoulder abduction
D_2 : Length from the original point A to the nut of the shoulder flexion
E_2 : Length from the original point A to the nut of the shoulder abduction
B_1, B_2 : Distance of each heel size
LA, LB : Length of each connecting rod
M_t : Transfer torque by own actuator

Clearly, we have a maximum value at shoulder flexion (40°) as shown in figure 3.
Namely, the optimum motion of the shoulder arm could be established in the position of maximum torque in the center of the movement area of daily life.

MECHANICAL SPECIFICATION

The structure for this new type arm is shown in figure 4. This arm has five degrees of freedom. It is composed of a shoulder joint, a elbow joint, a forearm and a hand.

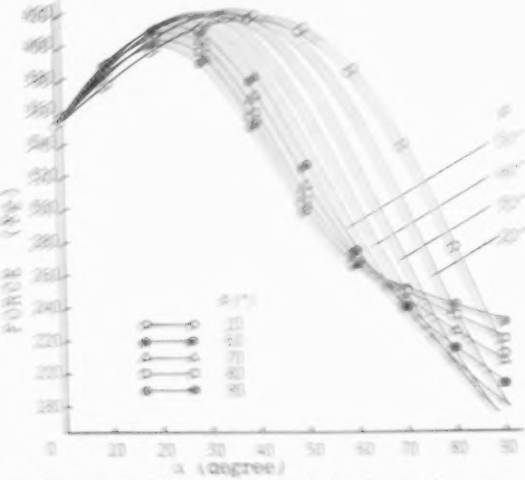

Figure 3 Resultant force of shoulder flexion

Table 1 shows these areas of motions and the weights of screws, gears, shafts, etc. are included in the weight of each part. The frame is made of carbon fiber-F.R.P. to lighten it.

Table 1

BEAM	PART	MOVEMENT	DEGREE	WEIGHT	LENGTH
1	shoulder	flexion extension abduction adduction	115°-15° 115°-15°	1.2Kg	295mm
2	elbow forearm	flexion extension pronation supination	0°-120° 100° 180°	0.77Kg	280mm
3	fingers	open	120°	0.18Kg	94mm

CONTROL SYSTEM

In general, if each degree of freedom is independently powered by its own motor, then a control system can be developed to co-operative motions in a variety of ways. Figure 5 shows a block diagram of this control system which developed as the prosthesis controller of the various function generators. The prosthesis controller contains two one-tip microcomputers, memories, five motor driver I.C, the mechanical switch, and the interface for a limit switch.

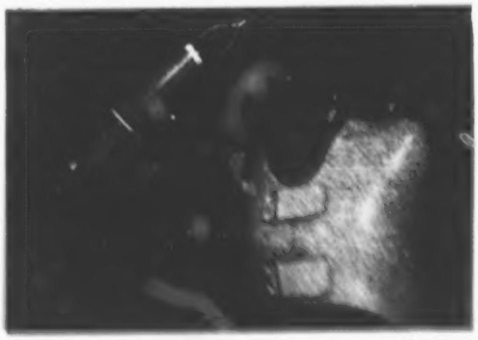

Figure 4 Shoulder prosthesis

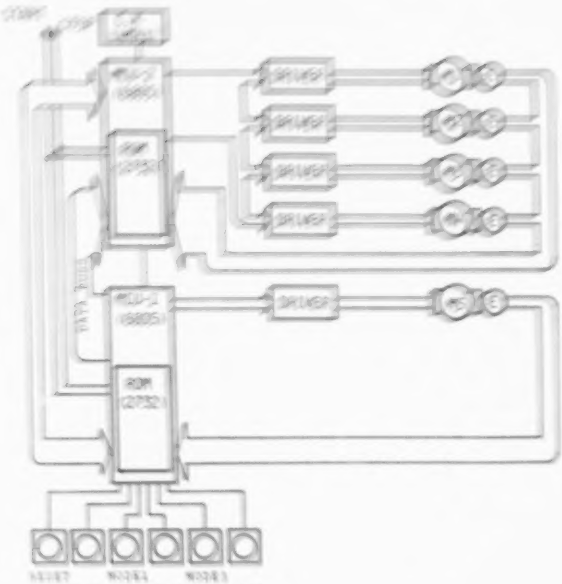

Figure 5 Control system

So that, this system is very small, simple and light. Two electric power surces are utilized; one is obtained from an alternating current for fixing and the other from Ni-Cd battery for transfer.

CONCLUSION

The feature of this prosthesis was to develop a double linkage mechanism of 2 degrees of freedom using a screw for the shoulder which had been one of problems. As we used the iron and aluminium, the weight became 2.1 kg. Therefore, the light weight is possible in case of using a carbon fiber etc. And then, the miniaturization became possible by using 6805 to the microcomputer units in controller.

We think that this is the new type prosthesis which lies in the middle of high functional prosthesis as a simple electrical arm.

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REAR CASTER WHEELCHAIR DIRECTIONAL INSTABILITY

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ABSTRACT

The directional instability of a wheelchair is shown to depend on the placement of the caster wheels and, consequently, the relationship between the center of gravity on the wheelchair and the fixed wheels. With the center of gravity in front of the fixed wheels (conventional arrangement), the wheelchair is directionally stable. However, when the center of gravity is behind the fixed wheels (rear caster arrangement), the wheelchair is directionally unstable, and the user must take corrective action in order to move in a straight line. A simple analysis for the control of a rear caster wheelchair is presented, and the fundamental variables of the problem are established.

INTRODUCTION

The problem of directional stability of a vehicle is not new, and there have been several papers on the subject of aircraft landing gear design in which the problem of directional stability while taxiing has been discussed. For example, in 1942 Jenkins and Donovan [1] observed that the distance of the main wheels aft or forward of the center of gravity is one of the most important dimensions of a landing gear. This design variable influences the directional stability, longitudinal stability, resistance to overturning, and loading of the wheels. They point out that a conventional (castered tailwheel) landing gear is unstable and that tail rudder control and wheel braking are necessary to avoid ground looping, whereas the tricycle landing gear is always directionally stable.

THEORY

In textbooks on the design of the airplane under the subject of landing gear layout, it is shown that directional stability of tailwheel versus nosewheel caster wheel placement depends upon the position of the center of gravity of the vehicle with respect to the fixed wheels. For example, Stinton [2] shows that the aircraft has directional stability on the ground when the center of gravity is ahead of the fixed wheels (tricycle). In the case of a conventional wheelchair as shown in Figure 1, an automatic correcting moment is developed by the fixed wheels which causes the wheelchair to return to rolling straight ahead when it happens to be deflected sideways by the road surface. The wheel force F_n has been found to be proportional to the angle θ, e.g., see Moreland [3]. The distance A which is measured perpendicularly from the fixed wheels axle to the center of gravity of the wheelchair depends on the loading and wheel layout. Translating $2F_n$ to the center of gravity gives a resultant force $2F_n$ and moment $M = 2AF_n$, where the moment is in the direction to cause the wheelchair to return to the line of motion.

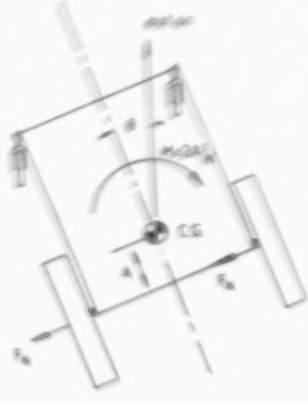

FIGURE 1 STABILIZING MOMENT FOR COVENTIONAL WHEELCHAIR

For a rear caster wheelchair as shown in Figure 2, the moment at the center of gravity due to the front fixed wheel forces $2F_n$ is in a direction which will cause the wheelchair to turn away from the intended direction of rolling. Thus, the wheelchair is directionally unstable, and it is necessary for the user to exert a corrective force on the hand rim or apply corrective torques with the joystick of an electric wheelchair controller. From this simple description of the problem, it is obvious that reducing A as much as possible will make the rear caster wheelchair easier to control but not eliminate its directional instability.

CONTROL PROBLEM

It is interesting to look at the rear caster wheelchair control problem from a dynamic point of view in order to determine the important control variables. The problem in its most elementary form involves a rotating mass with a polar moment of inertia, J, driven by a moment $M = 2AF_n = 2AC_0 = C_1\theta$. Assuming that the wheelchair is turned away from its line of motion by a bump in the road to θ_0 and no corrective action is taken for a time t_1,

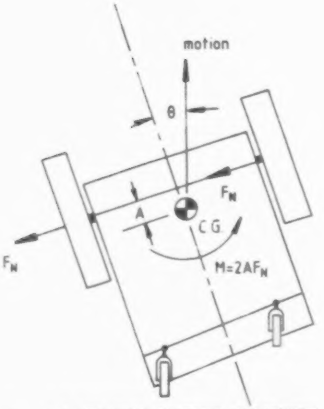

FIGURE 2 DESTABILIZING MOMENT FOR REAR CASTER WHEELCHAIR

the differential equation of the system and resulting angular motion to time t_1 is shown below.

$$J\frac{d^2\theta}{dt^2} - C_1\theta = 0 \tag{1}$$

The solution to (1) is:

$$\theta = \frac{\theta_0}{2} (\exp((C_1/J)^{\frac{1}{2}}t) + \exp(-(C_1/J)^{\frac{1}{2}}t)) \tag{2}$$

At time t_1, the angular position, velocity and acceleration of the wheelchair is found from (2).

$$\theta_{t_1} = \frac{\theta_0}{2} (\exp((C_1/J)^{\frac{1}{2}}t_1) + \exp(-(C_1/J)^{\frac{1}{2}}t_1)) \tag{3}$$

$$\dot{\theta}_{t_1} = \frac{\theta_0}{2}(C_1/J)^{\frac{1}{2}}(\exp((C_1/J)^{\frac{1}{2}}t_1) - \exp(-(C_1/J)^{\frac{1}{2}}t_1)) \tag{4}$$

$$\ddot{\theta}_{t_1} = (C_1/J)\theta_{t_1} \tag{5}$$

At time t_1, it is assumed that the user applies a restoring torque T to the wheelchair through the fixed wheels. Now the differential equation is given as follows.

$$J\frac{d^2\theta}{dt^2} - C_1\theta = -T \tag{6}$$

The initial conditions are given by equations (3) and (4) and the solution to equation (6) is

$$\theta(t') = \tfrac{1}{2}(\theta_0\exp(t_1/\tau) - (T/C_1))\exp(t'/\tau) + \tfrac{1}{2}(\theta_0\exp(-t_1/\tau) - (T/C_1))\exp(-t'/\tau) + T/C_1 \tag{7}$$

where t' is now the time beginning from the delay time t_1 and τ is the time constant of the system

$$\tau = (1/(C_1/J))^{\frac{1}{2}} \tag{8}$$

Clearly it is intended that $\theta = 0$ in some time t'/τ, which gives

$$T/C_1 = \frac{\theta_0\exp((t_1+t')/\tau) + \theta_0\exp(-(t_1+t')/\tau)}{\exp(t'/\tau) + \exp(-t'/\tau) -2} \qquad (9$$

$$T/C_1 = m\theta_0\exp(t_1/\tau) \tag{10}$$

A plot of equations (2) and (7) for a control delay time equal to one time constant and for two magnitudes of restoring torque, i.e., $T/C_1 = m\theta_{t1}$ where m = 2.2 and 2.04, are shown in Figure 3. The plots assume that the restoring torque is constant whereas one would decrease the restoring torque (shown as T→0) when the wheelchair turns toward the line of travel in order not to overshoot the straight ahead position. Since there will be other inadvertent deflections of the wheelchair due to the terrain or other interaction, the control problem keeps repeating itself and the user will soon find this directional instability problem extremely fatiguing. Automatic controls in an electric wheelchair's controller can be used to overcome the problem.

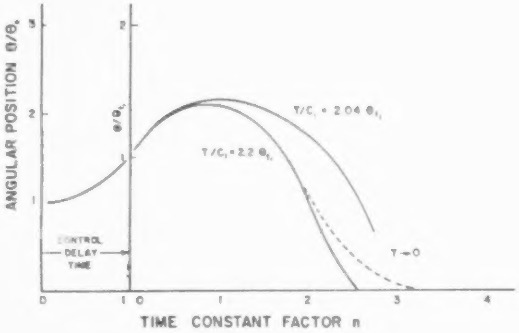

FIGURE 3 REAR CASTER WHEELCHAIR CONTROL CHARACTERISTICS

CONCLUSION

It has been shown that a rear caster wheelchair has an inherent directional instability problem. The controls necessary to overcome the instability are seriously debilitating for the user of a manual wheelchair but are feasible for an electric wheelchair. The analysis indicates a need for an experimental study of the problem, and results of such a study will be presented in a future paper.

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EFFECTS OF SIDE SLOPE ON WHEELCHAIR PERFORMANCE

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ABSTRACT

Compensation for the downhill turning moment of a WC on a 2-degree side slope results in a retarding force approximately equal to the rolling drag of a WC on a level surface. The total drag force on the WC while traversing a sloping surface is, therefore, roughly double the rolling drag. In contrast, the net energy cost of propulsion on this side slope is only 30% greater than for a level surface. Side slope propulsion is managed by "dragging" the uphill rim while pushing the downhill rim. Although this results in increased mechnical efficiency through greater use of a smaller muscle mass, it is more difficult and tiring for the user.

INTRODUCTION

The relative ease (or difficulty) in propelling a wheelchair (WC) with handrims is dependent on a number of factors. Included among these are such variables as the weight, physical dimensions, and materials of the WC, the physical dimensions and capacities of the user, the compatibility of WC and user dimensions, and external factors such as the texture, hardness and slope of the surface upon which the WC is operated. All of these factors have been elaborated to some degree by a number of investigators. Despite impressive efforts by various investigators to quantify the effects of these different factors there remain a substantial number of significant problems. Much of the difficulty in improving WC performance is a result of the interaction of the variables mentioned above and the fact that improvement with respect to one factor often results in undesired changes in others.

One problem of nearly universal applicability is the downhill turning tendency on sloping surfaces. This results from the characteristic mass distribution of a WC and occupant relative to the wheel orientation and the fact that nearly all outdoor, improved surfaces (e.g.,streets and sidewalks) are sloped for drainage. It is arguable whether this phenomenon is the most significant problem in WC mobility; however, it was identified as such in a National Report on technology for the handicapped. (1) While most WC users and others familiar with problems of WC mobility would likely agree that this is a problem of significance very little has been accomplished beyond identification of this effect as a problem.

The Problem

The purpose of the present investigation was to identify and quantify the factors affecting direction stability of manual WCs on uneven and sloping surfaces and to recommend potential means of controlling WCs.

METHODS

Drag forces were determined for two different WCs at 3 and 4 km/hr on a motorized treadmill with the bed level and with the bed inclined laterally at 2 degrees. These measurements were determined with the WCs tethered to a load cell attached to the front frame of the treadmill. The measurements were made with the test subject seated in the WCs. The WCs were "steered" for the slope condition by having the subject apply a resistive force to the uphill handrim.

	0 DEGREES				2 DEGREES			
	"STANDARD" WC		"SPORT" WC		"STANDARD" WC		"SPORT" WC	
Speed km/hr	3	4	3	4	3	4	3	4
Drag N	7.60	7.96	6.62	6.98	15.97	15.29	14.70	13.48
Power W	6.38	8.85	5.51	7.76	13.31	16.99	12.25	14.97
Strokes/min	41	44	41	45	42	47	42	46
Heart Rate	88	91	89	86	103	106	96	100
VO2 net L/min	.24	.30	.21	.28	.34	.37	.27	.35
Energy Cost W	83.7	104.63	73.24	97.66	118.59	129.05	94.17	122.07
Mech. Eff. %	7.62	8.45	7.52	7.95	11.22	13.17	12.61	12.26
VO2/Dist. L/km	4.8	4.5	4.2	4.2	6.2	5.55	5.4	5.25

*VO2 rest = .19 L/min, H.R. rest = 69 beats/min

TABLE 1. Wheelchair Drag and Propulsion Data for Level and Sloped Surfaces.

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Each of the above conditions was repeated with the subject propelling the WCs at the pace set by the treadmill. The subject's oxygen consumption, heart rate, and stroke rate were determined while he propelled the WCs. The exercise bouts were conducted over 5-minute periods with the measurements determined for the 4th and 5th minutes.

The subject was an athletic 20-year old male paraplegic (T12,L1). The WCs used were a "standard" model with the rear axle located on the rear vertical frame member and a "sport" model with adjustable axle. The axle position used for the latter WC was 2.5 inches in front of the rear vertical frame member. This WC also had a 3 degree camber in each drive wheel.

A static analysis of the 2 degree slope condition was made to determine the downhill turning moment to provide a basis for comparison with the results of the drag tests.

RESULTS

The results from the drag tests and exercise tests are presented in Table 1. An inspection of this table reveals that drag was approximately 12% higher for the "standard" WC. Drag was higher for the 4 km/hr condition on the level surface but was higher for the 3 km/hr condition on the sloped surface for both WCs. Further inspection of Table 1. reveals that the drag was roughly two times as large for both WCs at both speeds on the sloped surface as it was on the level surface.

The physiolgical effort required for the various conditions is reflected by the respective oxygen consumptions and heart rates. Both of these values were ordinally consistent with the power requirements determined for the different conditions; however, it can be seen that the oxygen consumption increased by only about 30% from the level condition to the slope condition while the power required to propel the WCs increased more that 100% with respect to these conditions. These differences correspond necessarily with the mechanical efficiencies for the various conditions since the percent efficiency is based on the ratio of power required to energy cost. According to this definition, efficiency of propulsion on the sloped surface is higher. When "efficiency" is interpreted as the ratio of net oxygen consumption (or energy cost) to distance traveled it can be seen that propulsion on the level surface is more "efficient".

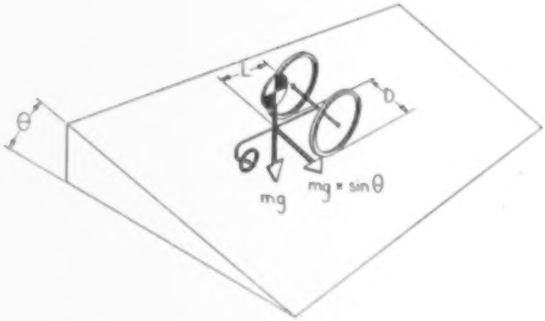

Figure 1: Static Analysis of Downhill Turning Moment

The factors which produce the downhill turning effect are identified in Fig. 1. These factors include the slope (θ), the moment arm of the c.g. about the downhill wheel (l), the mass of the WC and occupant (m) and the distance between the wheels at the surface (d). Therefore,

downhill turning moment = $mgl \sin \theta$, and,

drag on uphill wheel = $(mgl \sin \theta)/d$.

If $l = .15$ meters, $m = 80$ kg., $d = .56$ meters and $\theta = 2$ degrees, then,

downhill turning moment = 4.1 N.m.,

and,

required drag for uphill wheel = 7.3 N.

The drag for a WC with the above dimensions on a 2 degree slope would be the sum of the rolling drag and the drag necessary to counter the downhill turning moment (i.e., rolling drag + 7.3 N.).

	"Standard" WC		"Sport" WC		
SPEED	3	4	3	4	Ave.
(N)	8.37	7.33	8.05	6.50	7.57

TABLE 2. Differences in Drag Forces Between Level and Slope Conditions

The respective differences in drag determined for the level and 2 degree slope conditions are presented in Table 2. It can be seen that the average difference for the four conditions is 7.57 N. It can also be seen that the differences are less for the 4 km/hr condition for both WCs.

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DISCUSSION

An increase in drag due to the side slope was anticipated; however, the magnitude of this increase was somewhat surprising. The correspondence of the experimental results with the predicted value based on static analysis would appear to confirm the accuracy of these measurements. The higher drag values at the 3 km/hr speed for the slope condition is attributed to more frequent and higher amplitude oscillations from the line of progression. This was evident from the analog recordings of the forces with respect to the different speed conditions and was also consistent with subjective observations.

The higher mechanical efficiency obtained for the side slope conditions may be attributed to the more favorable conditions with respect to the force-velocity relationship of the muscles and also to the fact that only one arm was utilized for propulsion. The significance of the latter is in part a result of having only one arm active in the recovery phase. The recovery phase typically accounts for 75% of the stroke time and consumes metabolic energy but does not produce any work. This explanation is reinforced by the fact that the stroke frequency was nearly constant for all test conditions.

When the ratio of energy cost to distance traveled is used as the efficiency criterion it can be seen that propulsion on the level surface is more "efficient". This measure of efficiency is also consistent with the perceived effort by the subject for the different conditions.

It is evident from both the static analysis and the experimental results that the downhill turning moment and, correspondingly, the power requirement are decreased by moving the rear axle position closer to the c.g. and increasing the effective wheel width dimension with camber.

Two potential design solutions to eliminate the side slope effect are the c.g. WC (c.g. positioned over the drive axle with casters in front and back) and "steerable" casters. Although a properly balanced c.g. WC eliminates side slope effect it also eliminates the directional stability or tracking tendency of the WC and, therefore, requires nearly constant steering corrections. An acceptable "steering" mechanism for casters would not require manual control or it would be self-defeating. This effectively reduces the options to a weight shift mechanism.

Two different concepts based on weight shift to control the casters are under consideration at the UVA-REC. A design by McLaurin and Stapleton which utilizes a novel suspension mechanism on a 3-wheel undercarraige has been developed to the prototype stage. This design works quite well but would be difficult to adapt to a 4-wheel configuration.

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6.3

STRESS ANALYSIS OF WHEELCHAIR FRAMES

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ABSTRACT

In previous stress analyses, wheelchair frames have been rigidly modeled with uniform load distributions. The non-uniform load distribution in this analysis was determined experimentally. A three-dimensional, finite element model was developed which modeled the folding capability of three commercially available manual wheelchair frames. After further experimentation, the stress distribution determined with the finite element model was verified experimentally. From analyzing the stress distributions and geometries of the three frames, design recommendations are made.

INTRODUCTION

In the literature, studies exist on the application of computer aided techniques to the design of tubular wheelchair frames [1-2]. These studies describe either static, modal or transient analysis. Results of these studies usually indicate that the wheelchair frames are sub-optimally designed with respect to strength and that reduction in weight could be accomplished by redesign. These analyses have ignored two important technical points: 1) the accuracy of the loading schemes and 2) the accuracy of the three-dimensional model. The loading scheme most commonly used has been a uniformly distributed load over the sling-seat supports. The three-dimensional models used were rigid frames that did not allow for the joint release which represents a foldable design. Both of these points are discussed in this paper.

Experimental tests at the University of Virginia Rehabilitation Engineering Center have shown that the actual loading is not uniformly distributed over the sling seat supports. The experimental technique used to measure the actual loadings is discussed. Using the experimentally determined loads, a computational three-dimensional static stress analysis was performed on models of three commercially available, manual wheelchair frames. With additional experimentation, the models were verified. The ANSYS computer code, a general purpose finite element package, was used for the analysis. Results from rigid and foldable models of wheelchair frames are compared. The onset of yielding is also discussed. By analyzing the similarities and differences of the computed stress and strain data and wheelchair geometry, recommendations are made for future designs.

METHOD

1. Wheelchair Frame Loading

A Stainless Sport wheelchair was instrumented with strain gages to measure the strains produced when the chair was loaded with both live subjects and test dummies. The live subjects weighed 142 and 170.5 lb_f with respective heights of 66 and 73 inches. The test dummies consisted of the ISO standardized dummy [2] and a 72 kg. anthropometric dummy.

The sling-seat and seat back of many folding wheelchair designs are attached to the frame by screws. The loads due to an occupant in the seat are transmitted as applied forces rather than distributed loads at these screw locations. The loads were assumed to be equal on the left and right sides of the wheelchair. Due to the catenary nature of the seat fabric, each force is directed at an angle measured from the horizontal. Each force produced a torque on the frame which was measured with a strain gage bridge. Torque data was reduced to determine the in-plane forces at the screw locations. Axial loads at the screw locations were measured with additional strain gages.

The load on the footrest plate due to an occupant was resolved into a single force vector acting normal to the plate surface. By measuring two bending moments on the footplate support tubes and using the angles of inclination of the foot plates, the loading was determined.

2. Computer Model

An accurate finite element model of a wheelchair frame is needed to predict the structural response of the frame geometry to particular load inputs. A finite element model of a Stainless Sport wheelchair containing 220 nodes and 202 elements was developed for use with ANSYS, a large scale, general purpose, finite element computer package. Models of an Everest and Jennings Premier wheelchair and an Invacare Series 900 wheelchair were also developed.

Four types of elements were used in the models: three-dimensional beam elements, three-dimensional pipe elements, rectangular shell elements, and laminated shell elements. At several locations on the wheelchair frames, there existed physical situations that could not be modeled by simply connecting elements together (i.e., a tube within a tube, rotating tubes, etc.). These situations exist due to the folding capability of the frame. Specific areas given special attention were seat back tube attachments, back seat corners, front seat corners, and bottom horizontal tube connection to the cross-brace, Figure 1.

The wheelchair models were constrained at the front caster axles, the rear wheel axles and at the cross brace connection bolt. The front caster axles were constrained in the X-coordinate and Y-coordinate directions. The rear wheel axles were constrained in the Y-direction. The cross brace connection pin was constrained in the Z-direction and also constrained not to rotate about the X-axis. The global coordinates are shown in Figure 2.

The wheelchair computer model was statically loaded at the seat screw locations with the experimentally determined values. Node displacements and element stresses were determined using ANSYS.

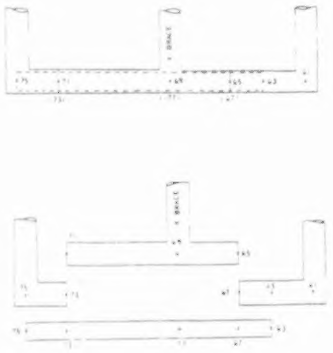
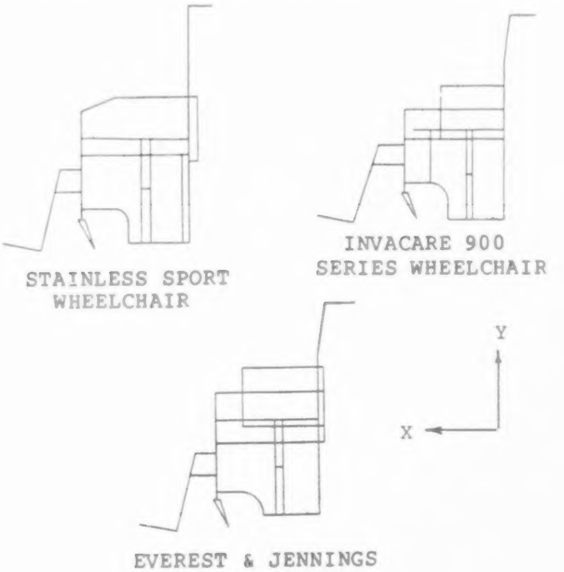


FIGURE 1 Lower Horizontal Tube Assembly



STAINLESS SPORT WHEELCHAIR

INVACARE 900 SERIES WHEELCHAIR

EVEREST & JENNINGS PREMIER WHEELCHAIR

FIGURE 2 Side View of Three Frame Models

RESULTS

The accuracy of the computer model was verified experimentally. Strain gage rosettes were mounted on the Stainless Sport wheelchair at the cross tubes and the area behind the caster forks. Computational and experimental principal stresses are shown in Table 1. The difference between the two types of stresses was less than 20% in all cases.

TABLE 1

NODE	Model (psi)	Experimental (psi)
65	12,452	11,525
68	9,563	10,511
69	9,295	8,973
72	12,002	10,488
103	2,276	2,505

With the wheelchair occupied, the percentage of total body weight on the footplate and seat screws is shown in Figure 3. The areas experiencing the highest stress due to static loading included the horizontal seat tubes and the crossbars on all three wheelchair frames. This part of the frame supports and transmits a majority of the torsional and bending loads to the rest of the frame. Stresses in this area reach a maximum at the cross brace bolt location in all three wheelchair designs.

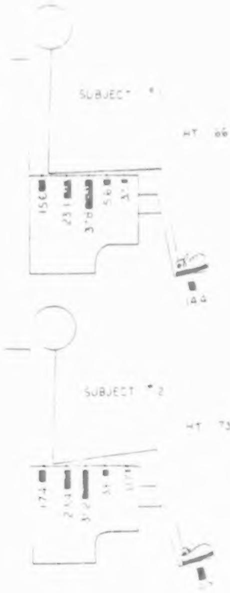


FIGURE 3 Wheelchair Seat Load Distribution

Establishing yielding as a criterion for failure, a safety factor can be calculated for each node location using the distortion energy theory of failure. The principal stresses generated by the computer model on the cross tubes are given

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in Table 2. A safety factor based on the yield strength of UNS S30400 stainless steel (Yield Strength = 35,000 psi) is also listed.

TABLE 2

FRAME MODEL	MAXIMUM STRESS (PSI)	SAFETY FACTOR
E & J	22,097	1.58
INVACARE	21,561	1.62
STAINLESS	22,859	1.54

Although the factors of safety are approximately one and a half, these values consider only static loading. Also, there exists on the three frames a transverse hole in each of the crossbars at the bolt location. The presence of these discontinuities magnifies the stress distribution.

Values of the stress concentrations are dependent upon the type of loading and the geometry of the discontinuity. The stress concentration factors used with these particular computer models were K_t= 1.90 and K_{ts}=1.42. With these stress concentration factors, the safety factors were computed to be approximately 0.83. In all cases, the area around the bolt holes on the crossbars possesses a high likelihood of yielding due to static loading.

Maximum stresses on the cross tubes were computed for four variations of the Stainless Sports wheelchair model. The variations consisted of combinations of released and rigid joints and applied and distributed loading. Resulting stresses for each case are listed in Table 3.

TABLE 3

	Maximum Stress (psi)
Released Joints - Applied Load	22,859
Released Joints - Distributed Load	23,647
Rigid Joints - Applied Load	22,441
Rigid Joints - Distributed Load	22,312

Although most folding wheelchair frames are similar, differences in design exist. Side views of the three frames evaluated are shown in Figure 2. Armrest shape, front vertical tube placement and the crossbar connections vary among these three wheelchairs.

The differences among the armrests on the three frames did not greatly influence the response of the frame. Armrests merely supply a small degree of dimensional stability to the frame.

The front vertical tube on the Invacare frame is located further back than either the Stainless or Everest & Jennings models. This tube supports the second upper horizontal seat tube. Unlike the Everest & Jennings and Stainless wheelchairs, this tube provided no vertical support to the seat support side rails. An increase in the stresses experienced on the second horizontal tube was observed. The magnitudes of the stresses were over twice those found on the Everest & Jennings and Stainless frames at similar locations. The advantage of this support was that it decreased the stresses experienced on the area of the side frame adjacent to the front caster assembly. Stresses in this region were smaller primarily due to a significant reduction in the bending stress at the location.

The Stainless wheelchair has two vertical tubes on the back part of the chair with the rear wheel connection on the forward tube. The addition of the extra tube had little effect on the stress distribution when compared to the other frames.

The front crossbar on the E&J and Stainless frames connects the upper horizontal tube on the left frame to the lower horizontal tube on the right frame, while the opposite occurs on the Invacare frame. Due to the fact that one crossbar crosses in front of the other, an unequal load distribution occurred at the four wheel support points. The difference was approximately 15 percent of the total load.

From these preliminary results several design recommendations can be made. The primary area recommended for redesign is the center of the cross brace with an overall reduction of the tube wall thicknesses in the remainder of the frame. Because this component carries a majority of the seat loads and stress concentrations, this area is particularly prone to yielding. It is also recommended that the front vertical support for the seat tube be moved back and connected to the lower and second upper horizontal tubes. This serves to distribute the seat loading to less critical areas on the frame.

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Acknowledgement

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EVALUATION OF RIDER COMFORT WITH FRONT CASTER SUSPENSION

6.4

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ABSTRACT

Several front caster assemblies have been designed with energy absorption devices to prolong the structural component life of the casters. The effects of the suspension forks on ride firmness and rider comfort were investigated experimentally and with computer analysis. Advantages and disadvantages for each of the suspension systems were discussed.

INTRODUCTION

Over the past several years, designs have been presented to improve the life of front caster fork assemblies. Being the first part of the wheelchair to strike objects in its path, the front caster assembly is particularly prone to damage. The common feature of new designs is the inclusion of an energy absorption device. With this device present, the damage to the structural components of the caster is reduced upon collision with obstacles. The purpose of this paper is to discuss the effects of these suspension forks on wheelchair ride quality.

A regular stiff fork and three suspension forks were studied. The physical characteristics of the suspension forks were determined experimentally. These characteristics were used to evaluate ride firmness and rider comfort both experimentally and with a computer model. Then, ride firmness and rider comfort were used to determine the overall performance of each of the forks.

DISCUSSION

A regular caster fork and three suspension forks manufactured by Davis, Mulholland and Everest & Jennings were used during the investigation. The designs of the suspension forks vary with respect to pivot point placement and the type of energy absorption device used. The pivot points on all of the suspension forks are shown in Figure 1. The Davis fork uses a compression helical spring as its energy absorbing device. The Mulholland uses a beam and the Everest & Jennings uses pre-loaded, stacked leaf springs. A conventional fork neither pivots nor has energy absorption devices other than its structural components.

Static vertical load displacement curves are shown in Figure 2 for the three suspension forks. Vertical spring rates are similar, but the torsional spring rates vary as shown in Table 1. The variance in torsional spring rates is due to changes in the suspension pivot point location with respect to the wheel axle. For example, the Mulholland fork has a longer moment arm than the Davis fork, 64 mm compared to 23 mm. Hence, for the same applied vertical load, the resulting moment about the Mulholland pivot will be about three times greater.

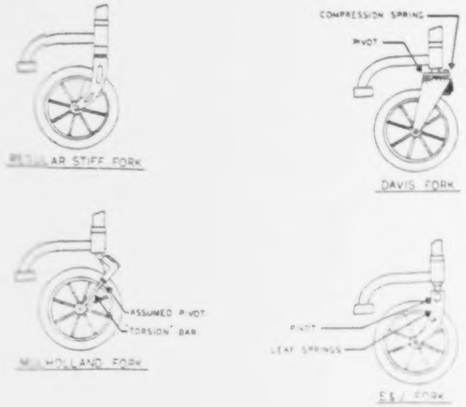

FIGURE 1 Caster Fork Designs

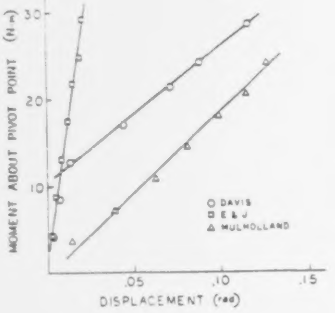

FIGURE 2 Moment Versus Fork Displacement

Table 1
FORK STIFFNESS

TYPE	K_{tors} (Nm/rad)	K_{vert} (kN/m)
Regular	753	212
Davis	153	93
E&J	1300	49
Mulholland	188	30

The helical compression spring in the Davis fork design allows the capability of changing the spring rate by changing the spring. This is a desirable feature because different springs could be prescribed for different rider weights to obtain the best suspension. However, this fork design does not allow for adequate clearance before spring clash occurs. Therefore the softer springs cannot be pre-loaded enough to support the static load without contacting the stops.

The leaf spring incorporated into the Everest & Jennings fork permits variation in the spring stiffness by reducing the number of leaves. This fork uses five stacked, pre-loaded leaf springs. The major disadvantage of this design is that only 8 to 10 degrees of rotation are allowed between the stem and pivoting caster support arms which severely limits the suspension travel.

A finite element model developed with the ANSYS computer code for static analysis of the 900 Series Invacare wheelchair was modified for modal and transient analysis. The computer model of the wheelchair does not include the sling seat or occupant. Masses on the seat and back tubes represent the wheelchair loading. Since the dynamics of the rider are not included in the model, resultant frequencies are lower than those obtained experimentally.

Stout expressed ride firmness in terms of the minimum value for the system's lowest natural frequency (1.6 Hz.) [2]. Using this criteria, none of the three suspension forks mounted on the Invacare Series 900 frame yielded a ride too soft. Experimentally, the lowest natural frequency was determined to be 5.5 Hz. Using the computer model, modal analyses were run for each of the suspension forks with the results shown in Table 2. The lowest natural frequency from the computer simulation was 6.5 Hz.

Evaluation of rider comfort is based on the magnitude of the acceleration of the center of gravity (CG) of the wheelchair. The standard equation for comfort was determined with data collected by Jacobson [3] and correlated using the technique given by Thacker [4]. Vertical acceleration of the wheelchair frame CG was determined with computer analysis and with experimentation. Plots of the CG acceleration for a wheelchair traveling over a 5/16 in. bump are shown in Figures 3 through 8.

Since the dynamics of the rider were not included in the computer model, differences exist between experimental and computational results. Results from the computer model with a 64 kg load and damping more closely matched the experimental data from the wheelchair loaded with 45 kg sandbags as shown in Figures 4 and 6. Experimental data with the 72 kg anthropometric dummy had peak accelerations that were twice the magnitude over the computational results. If some damping was present in the caster pivot, the magnitudes of the peak accelerations increased. Large loads were less dependent on wheelchair speed than small loads.

Table 2
NATURAL FREQUENCIES OF SUSPENSION FORKS

Mode	Frequency (Hz)	
1	7.62	STIFF FORK
2	10.51	
3	13.83	
4	14.59	
5	18.48	
1	7.29	DAVIS FORK
2	8.39	
3	12.45	
4	13.68	
5	18.04	
1	7.27	E & J FORK
2	8.37	
3	8.85	
4	11.54	
5	17.43	
1	6.51	MULHOLLAND FORK
2	7.27	
3	8.33	
4	10.35	
5	17.60	

From the experimental testing of the caster forks, only the Mulholland fork improved rider comfort as determined by wheelchair frame CG acceleration. However, the undamped natural frequency of the loaded Mulholland fork is approximately the same as the first natural frequency of the wheelchair frame. This resulted in extreme suspension travel and bouncing of the wheelchair at a speed of 2 kph on flat surfaces.

The Davis and Everest & Jennings forks can be modified to increase rider comfort. Insertion of an elastomeric material at the stops will absorb some of the shock when the forks reach full deflection.

When the natural frequency of the front suspension is reduced, the ride is softened and larger suspension travel occurs. Below a natural frequency of about 10 Hz, the resulting frame CG acceleration over the bump was increased at speeds between 2 and 3 kph. Under these conditions, the wheelchair frame is approaching resonance. The frequency of the one cycle forcing function of the bump is near the natural frequency of the wheelchair. At higher speeds, the frequency of the forcing function was higher than 10 Hz resulting in less frame CG acceleration.

CONCLUSION

The effects of the suspension forks on ride firmness and rider comfort were studied. From this analysis, it is recommended that further changes be made to all of the fork designs. As presently designed, only the Mulholland fork improves comfort. However, its natural frequency is approximately the same as the wheelchair frame which leads to excessive bouncing on a level surface.

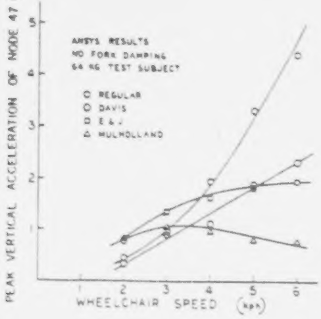

FIGURE 3 ANSYS CG Acceleration (no damping)

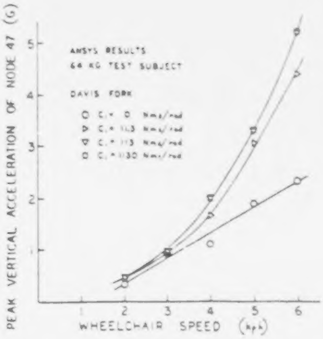

FIGURE 4 ANSYS CG Acceleration (variable damping)

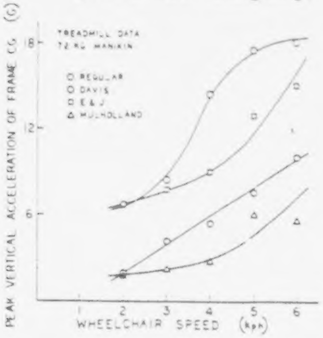

FIGURE 5 Measured CG Peak Acceleration

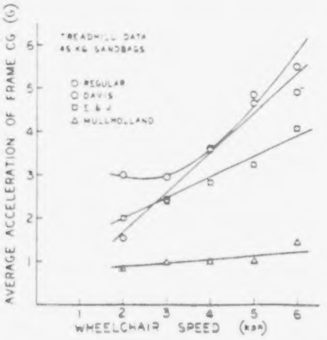

FIGURE 6 Measured CG Average Acceleration 45 kg Sandbag Load

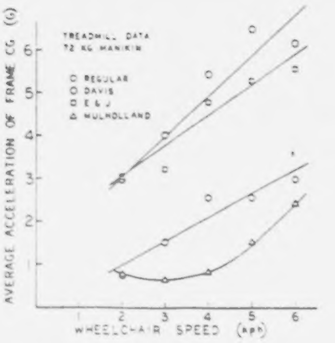

FIGURE 7 Measured CG Average Acceleration 72 kg Anthropometric Dummy

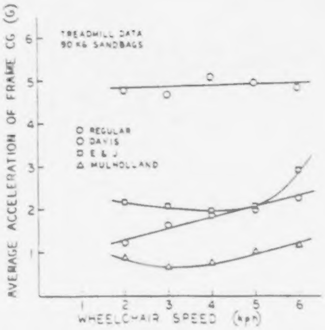

FIGURE 8 Measured CG Average Acceleration 90 kg Sandbag Load

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7.1

THE SCOLITRON AND THE L.E.S.S. TECHNIQUE: A MODEL FOR TRANSFER OF TECHNOLOGY AND INFORMATION

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ABSTRACT

In the mid 1970's, at Rancho Los Amigos Rehabilitation Engineering Center, a new method and associated equipment was developed for the treatment of scoliosis and kyphosis. To futher evaluate this method at other treatment centers it became necessary to involve industry in order to make equipment available to other investigators. In parallel, during the late 1970's, technology was transferred, patents were applied for, data forms were developed, clinician and patient manuals were designed, papers and exhibits were presented at conferences, a multi-center treatment program was established, an international advisory board was formed, and a computerized database was built. Later, in the early 1980's, patents were issued, FDA approval was obtained, several peer review papers were published, seminars were developed, and the public informed through the mass media. Presently, 220 centers domestically and 80 abroad are using the L.E.S.S. treatment originally developed at Rancho.

INTRODUCTION

At Rancho Los Amigos Rehabilitation Engineering Center, during 1976-77, transcutaneous electrical neuromuscular stimulation techniques were developed for the treatment of spinal deformities (1,2). Subsequently, patients with progressive idiopathic scoliosis who had refused any other form of treatment were accepted to the Lateral Electrical Surface Stimulation (L.E.S.S) program at Rancho Los Amigos Medical Center (3). Using a different electrode placement technique, but similar stimulation parameters, a group of kyphosis patients also began treatment (4). Portable single- and dual-channel muscle stimulators were developed at Rancho Los Amigos Rehabilitation Engineering Center. It quickly became evident that the REC was not set up to mass produce stimulators. The demand for stimulators at Rancho could not be fulfilled before commercial stimulators became available by the end of 1978.

TECHNOLOGY TRANSFER

Successful transfer of an invention or a new piece of equipment from a research institution to the market place is most often impeded by poor planning and insufficient communications between the parties. Particularly, when also a new concept must be proven scientifically, does technology transfer become very complex and involved. It was therefore felt that industry should become included already during the early phases of the L.E.S.S. project, at a time when we from a research standpoint were comfortable with the treatment results.

Contrary to popular belief, there is much more to technology transfer than just handing over blueprints of prototype equipment to a manufacturer. First a company must be sought which not only has the interest but also existing capabilities and capacities to undertake further development and sufficient production. There should also be a match between the type of proposed rehabilitation equipment (electronic, mechanical) and the type of devices currently fabricated by the company. Secondly, the manufacturer must provide a market plan including expected sales and channels of distribution. This forces an assessment of the validity of the technology transfer.

When, as in our case, the equipment is to be used for a new treatment method, the Food and Drug Administration (FDA) requires a multi-center evaluation of the efficacy of the device. Since the company producing the device is responsible for data gathering and final submission for pre-market approval to the FDA, a strong commitment must be made on behalf of the company. Such an investigation may easily be many times more expensive than just bringing a new product to market and the manufacturer must allow for such expenses in his long-range budget. To make the sponsoring of this investigation more attractive for the device producer it is suggested that the research organization file for a patent on the equipment and treatment method (if patentable) and grant the company an exclusive production license. In turn, the research institution may be able to accept a royalty for each unit sold which again could support further research in the field.

A close, coordinated collaboration must be established between the research organization, which originated the technology, and the manufacturer to assure that the equipment reaches the market place in a timely and uncompromised manner. Working as a team is advantageous to both parties. The researchers are able to expand their studies, thus assuring statistical significance of their material necessary for major scientific publications. The manufacturer is guaranteed support by experts in a field that may be somewhat unfamiliar to him.

EQUIPMENT AND MANUALS

As part of the technology transfer it is critical that a prototype of the manufacturers first device generation be made available to the equipment originator before the device is brought to market. In our case, laboratory test of several prototype units revealed that the specifications were not met to our satisfaction. These prototype were never applied to patients but returned to the manufacturer for improvements. Several versions were produced before an acceptable unit had been produced. This delayed production half a year, however, a costly major recall was prevented.

The ScoliTron, as the final product was named, was used to gather the required investigational treatment data for the FDA. To provide flexibility in selecting treatment parameters for the early clinical trials at Rancho, four screwdriver adjustable potentiometers were hidden beneath a trap door. The potentiometers came factory preset according to the protocol parameters selected for the multicenter study, but nevertheless, often both clinicians and patients alike accessed and maladjusted these parameters. It was learned that a medical device intended for home use should be as simple and foolproof as possible, preferably only having an ON/OFF button. For the second generation ScoliTron it was therefore decided to fix these parameters internally and only provide adjustment for the intensity of stimulation, which varies from patient to patient (Figure 1). After FDA approval of the single-channel unit in January 1983, a dual-channel ScoliTron was released for added treatment possibilities. Even that this unit was identical to the single-channel unit except for the added channel, a separate FDA data submission was required. Due to the amount of data requested, this multi-center investigation is still ongoing. Any institution contemplating transferring technology for medical treatment must realize that even minor changes in equipment specifications may require a separate multi-center investigation and FDA resubmission.

The most time consuming element of the technology transfer process is the writing of protocols, clinician manuals, patient manuals and data forms. This is best done by the research institution while editing, layout and printing may be done by the company which then carries the costs.

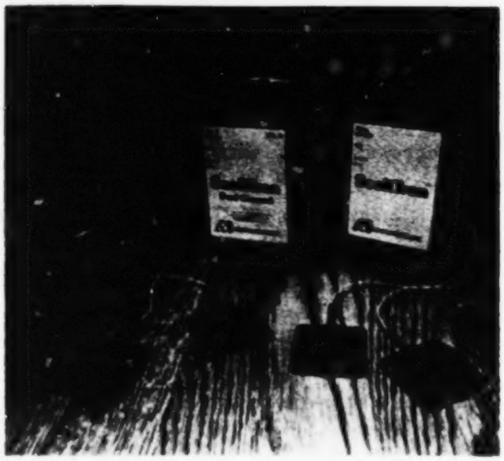

Fig 1. Single- and dual-channel ScoliTron manufactured by Neuromedics, Inc. A separate interrogator, not shown, will detect how many hours the devices have been used at the prescribed amplitude level.

INFORMATION DISSEMINATION

The dissemination of knowledge has taken different avenues. Since 1977, traditional paper presentations have been given annually at national and international conferences on scoliosis and kyphosis, orthopedics, rehabilitation engineering, etc. Scientific exhibits have been shown twice at the annual meeting for the American Academy of Orthopedic Surgeons and one time at the annual Scoliosis Research Society meeting. Upon FDA approval, commercial exhibiting has also been allowed. Presentations and exhibits at annual conferences are important for dissemination and exchange of current information. Unfortunately, these presentations are largely forgotten after only a few years and it becomes necessary to publish more substantial scientific papers in the major medical journals (1,3,4,5).

Teaching of the L.E.S.S. technique of treatment is accomplished through 1-day seminars followed by a clinic in-service provided by trained physical therapists (Figure 2). In addition, each clinic receives protocols, manuals and slide/sound presentations which they may keep for reference. To facilitate good patient/parent education, video tapes have been produced solely for that purpose.

Due to the novelty of the ScoliTron/L.E.S.S. providing an alternative to the body brace, the mass media (national radio and TV news, weekly magazines, family journals, etc., etc.) has focused on this treatment modality on numerous occations. This has increased the public awareness immensely. It must be pointed out, however, that one should be very careful with the press in not making overstatements regarding the significance of the research. Three hours of filming easily projects a distorted view when condensed to a 5-minute segment on the news.

Fig 2. Teaching the L.E.S.S. technique.

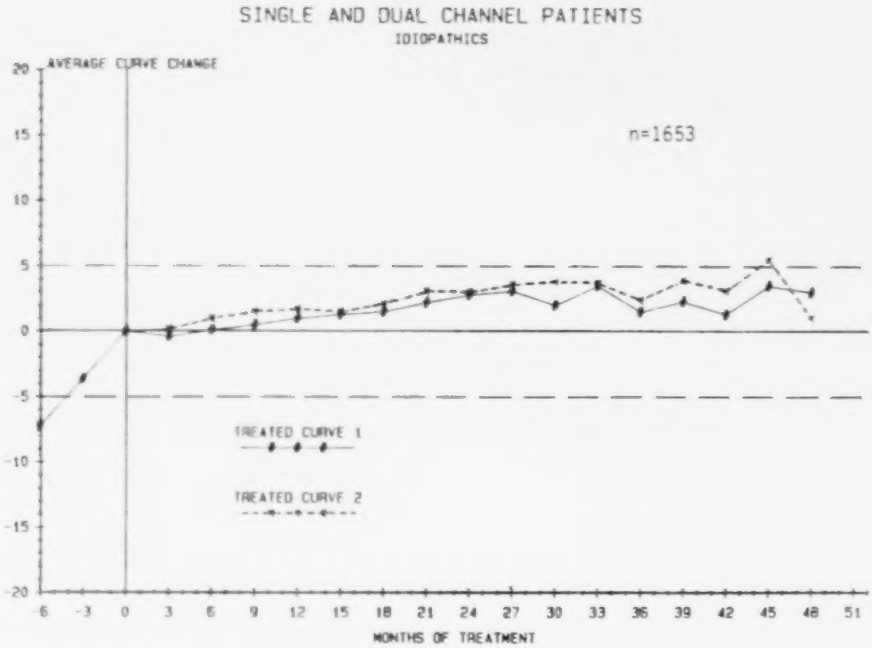

Fig 3. Average change in scoliosis curvature prior to and during time of treatment.

MULTI-CENTER STUDY

Prior to launching the multi-center clinical investigation of the ScoliTron/L.E.S.S. treatment method an international advisory board was selected from the membership of the Scoliosis Research Society (5). The advisory board reviewed the protocol and data collection procedures as proposed by the Rancho chief investigators and made recommendations for patient selection, treatment termination, database management, investigator selection and proper conduct of the multi-center study itself. This impartial advisory board met semiannually to review progress of the study and discuss data summaries from a database maintained by the manufacturer. This database was used to collect all data on all patients in the study and provide statistical analysis of the results. Although data submission to the database is not necessay for the FDA approved single-channel use, the database is still maintained. Any investigator may request a separate analysis of how well his clinic is doing in comparison to the average results of the multi-center study.

The present treatment outcome of 1653 patients entered into the database per December 31, 1984 is shown in Figure 3. It is seen that the progression of the scoliosis curvature prior to treatment initiation is halted with application of the ScoliTron. On the average, both single and double major curves stop further progression during the 4-year treatment span depicted.

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TESTING TECHNIQUES FOR THE COMMERCIALIZATION ADVANCEMENT OF ASSISTIVE DEVICES

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ABSTRACT

EIF/REC is researching the incentives and disincentives for the commercialization of assistive devices. The Center also is preparing to test alternative commercialization techniques to learn which are the more effective procedures for commercializing new assistive devices. This paper discusses some of the commercialization techniques to be tested, and the preparations now in progress for conducting these tests. Product developers, manufacturers, and marketers are invited to participate in the tests.

INTRODUCTION

Why aren't more new assistive devices commercially available to disabled people who could benefit from them? Technological advances are revolutionizing the lives of most of our society. But many believe these technological advances have not helped disabled people to the degree that they could.

The problem is not with the research and development of assistive devices. The National Institute of Handicapped Research (NIHR), the Veterans Administration, and others have established widely respected R&D centers for rehabilitation engineering. And many talented individuals have developed new assistive devices in their private shops. The problem is in advancing the prototype models from the R&D laboratories or private shops into production models, and thence into the hands of the people who can utilize them. This process is called "commercialization advancement".

In May of 1983, the Electronic Industries Foundation (EIF) established a Rehabilitation Engineering Center for Technology Evaluation and Commercialization Advancement through a cooperative agreement with NIHR. The Center is charged with development of techniques to increase the availability of assistive devices through the stimulation of private industry participation in the production, distribution, and marketing of such devices.

Consider the complete set of new assistive devices. Some of these devices are so exciting that they are available on the market almost immediately after development. A few others, unfortunately, may not be commercially viable products, and are unlikely ever to become commercially available. It is the devices between these two extremes which receive the greatest attention of the EIF Rehabilitation Engineering Center (REC). Many believe there are too many worthwhile devices in this middle category.

The EIF/REC is engaged in research of the incentives and disincentives for commercialization of assistive devices. The Center also is preparing for more pragmatic tests to compare alternative commercialization techniques. Through this testing of techniques, the Centers hopes to learn which are the more effective procedures for commercializing new assistive devices, and plans to develop recommendations for future commercialization activities.

This paper discusses some of the commercialization techniques to be tested, and the preparations now in progress for conducting these tests.

PRODUCT IDENTIFICATION AND SELECTION

The first step in preparing for any commercialization advancement process is the identification and selection of appropriate new assistive devices.

There are few limitations to the types of devices for which the EIF/REC wants to experiment. The Center is interested in devices which assist in overcoming any and all functional limitations which may handicap people with impairments. These devices can be useful for occupational, independent living, or recreational activities. Although the Center focuses mostly on assistive devices, clinical devices have been considered for commercialization experiments. EIF/REC is interested in electronic as well as any other type of assistive device (despite the name of the Foundation).

To consider a device for its commercialization tests, the Center has required that at least one working prototype model of the new device must exist. If not, the device idea is still in the research and development stage, and may be of interest to other RECs or R&D organizations.

Obviously, the developer of a new device and the EIF/REC must have expectations that a new device can be a commercially viable product. That means that a new device must be safe, and must serve some useful purpose for a significant number of disabled people. The device must be affordable for the prospective users, or there must be a possibility of third party financing. To limit duplication of effort and dilution of the market, the new device must differ from comparable devices already on the market. ABLEDATA is very useful for checking this.

Between February 1, 1984 and January 20, 1985, 43 devices came to the Center's attention for consideration. After some preliminary evaluations, 16 of these devices were selected as candidate devices, and 17 others were dropped from consideration. Decisions were pending for the remaining 10 of the 43 devices. In the coming year, the Center hopes to identify even more new devices.

The EIF/REC is documenting how each device comes to its attention, and the reasons for deciding whether to include a device in its tests. Such documentation may be helpful in preparing future recommendations.

Two common reasons for the Center's dropping a device are: (a) a working prototype model was not ready, and (b) the device was judged not to be commercially viable (for one of the reasons given above). In the case of problem (a), a device may well be reconsidered when a working prototype model becomes available. In the case of problem (b), EIF/REC is making a list of organizations and individuals who may be interested in new product ideas, and are not concerned about a product's commercial viability. An example of such an individual might be a rehabilitation engineer at a rehabilitation treatment center. With permission of the product developer, information still can be shared about a device which EIF/REC had to drop for reason (b).

PRODUCT PORTFOLIOS

The transfer of information about new devices to potential manufacturers and marketers is an important part of the commercialization advancement process. EIF/REC hopes to learn from its tests just what information is needed, and in what form it should be presented for efficient and effective results.

EIF/REC is beginning its tests by collecting information useful to the commercialization advancement process into "Product Portfolios". The contents of Product Portfolios may be modified during the course of the Center's tests, but for now they have three parts. "Part I: Device Characteristics" includes a brief description of the device, information about the estimated cost of the device, information about the training that may be needed for using the device, information about patents and licensing expectations, and information about any previous evaluations of the product. "Part II: Market Potential" includes an estimate of the identity and size of the target market(s). It also includes information about the possibility of third party funding, information concerning supply factors such as special production requirements, and information about any similar products already on the market. Information obtained from an ABLEDATA search is included here. "Part III: Other Considerations" contains any other type of information that needs to be considered for the commercialization of a device.

The EIF/REC recognizes that many Product Portfolios contain proprietary information which the developers/ inventors expect will be protected. For this reason, the Center has three "stages" for releasing information.

Stage I information contains no proprietary information, so it can be broadly distributed in, for example, mail-outs to prospective manufacturers of new devices. Stage I information is selected so as to catch the initial interest of potential manufacturers and marketers.

Stage II information is shared with a manufacturer after a non-disclosure agreement is signed. Stage II information includes most of the information contained in a Product Portfolio, but exclusivity of product is not promised to a manufacturer. That is, the product may still be offered to more than one manufacturer until Stage III.

Stage III status is given to a manufacturer for a product only after a non-disclosure agreement and perhaps a commercialization advancement plan have been received by the EIF/REC, and a licensing agreement has been reached with the developer/inventor. At this point, the product is no longer offered to other manufacturers.

MANUFACTURING AND MARKETING

As of this writing, the EIF/REC is just beginning to test and compare various techniques to enlist the interest of manufacturers and marketers in commercializing assistive devices. These techniques include: (A) Mailing Lists, (B) Individual Contacts, and (C) Vendor

Conferences. Other techniques which come to the Center's attention may also be included in the tests.

In the first technique mentioned above, (A) Mailing Lists, Stage I information about one or more new assistive devices is mailed to a number of manufacturers. To date, the Center has a list of over 800 manufacturers and marketers who may be interested in assistive devices. The list is still growing. EIF/REC's lists of new devices, manufacturers, and marketers are categorized by product types, manufacturing capabilities, and preferred marketing strategies (e.g., sales representatives, dealers, distribution networks, etc.). This categorization will allow more efficient mailings, and greater flexibility in testing commercialization techniques.

In the second technique mentioned above, (B) Individual Contacts, EIF/REC will attempt to enlist the participation of manufacturers and marketers through discussions with individuals in industry. When needed, the Center will have the assistance of the Electronic Industries Association and the Electronic Industries Foundation for making appropriate contacts within industry.

In the third technique mentioned above, (C) Vendor Conferences, manufacturers and marketers will be invited to attend a conference with product developers at which information is presented about one or more new devices. The manufacturer/marketer making the "best offer" wins the opportunity to commercialize the device.

One of these techniques may be more effective, for example, for devices which have utility to both disabled and able-bodied consumers. Or the effectiveness of a technique may differ for different types of devices (e.g., high tech / low tech devices). The Center's tests will be designed to investigate these and other possibilities.

After a product is manufactured, successful commercialization requires marketing strategies which move the new assistive devices from the producers to consumers. The target markets for most assistive devices, however, are "fragmented" markets. The number of potential consumers for a particular assistive device is often relatively small, and the consumers for that device are often scattered throughout the country. Manufacturers/ marketers of assistive devices have to cope profitably with this fragmented market.

The Center hopes to find effective methods for alleviating this and other marketing problems. One example of a possible countermeasure for fragmented markets --- marketing networks --- is already in used. An example of a marketing network is operated by the American Foundation for the Blind. The AFB catalog offers assistive devices from numerous producers, making possible certain economies of scale in marketing. EIF/REC will be developing recommendations concerning effective market strategies based on its commercialization tests.

CONCLUSION

Clearly, EIF/REC's research and commercialization tests can not succeed without participation from others. Therefore, the Center is asking for outside assistance.

If you are a developer of assistive devices, you are invited to submit your suggestions for devices for which you would allow the Center's collaboration via its commercialization tests. Proprietary information will be protected. Suggestions should be limited to assistive or clinical devices having at least one working prototype, which are not already commercially available. Please phone (202 955-5828) or write Robert Mills at the address given below.

If you are a manufacturer or distributor, the Center offers its testing activities as a means for liaison between your company and developers of rehabilitation aids. EIF/REC may be able to help you expand your product line to include the latest technologies for the disabled. Please contact Victor Knorr (202 955-5824) at the EIF/REC.

ACKNOWLEDGMENTS

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The Electronic Industries Foundation is a private, non-profit organization established in 1975 by the Electronic Industries Association to utilize the nation's talents and capabilities in electronics to the benefit of the general public and in the national interest. EIF seeks to engage industry in socially responsive programs.

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7.3

AN EVALUATION PROCEDURE FOR ADVANCING THE DEVELOPMENT OF REHABILITATION EQUIPMENT

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ABSTRACT

A procedure for engineering analysis and clinical evaluation of developmental, user-ready rehabilitation equipment is presented. The validity of the evaluation procedure and description of the process are illustrated by using the Stanford Storable Crutch as an example.

INTRODUCTION

Centralized, focused programs for evaluation of rehabilitation devices and methods have been few. Most notably, the National Academy of Sciences (NAS) and the Veterans Administration (VA) were the first to assume responsibility for clinical evaluation services, research coordination, and standards setting - activities that ultimately led to commercialization and widespread use of various rehabilitation products. Subsequent to the demise of these programs, evaluation activities became scattered among individual programs that performed certain aspects of this function, often on their own developmental products.

In mid-1983, the National Institute of Handicapped Research (NIHR) established two centers to deal exclusively with the issues of evaluation of rehabilitation technology and stimulation of industry. Created as a centralized, objective component of the Rehabilitation Engineering Center system, these Centers seek to identify appropriate developmental products and technological applications for increasing independence and employability of disabled persons, and then garner the resources to enhance their further development and commercialization. However, as experience has demonstrated, advancing a prototype device to the level of a potentially commercial product often requires that the device be guided through a comprehensive, cohesive evaluation process.

SwRI-REC EVALUATION PROCEDURE

A systematic plan for evaluation of prototype, user-ready rehabilitation devices was devised by the SwRI Center staff. Major elements of this plan include engineering test and analysis, preliminary user trials, and long-term clinical evaluation. Generally, information resulting from these activities would include product safety, durability, and reliability indications, operating characteristics, appropriateness of components and materials, identification and definition of appropriate applications, and user acceptance information. Portions of the plan were modeled from procedures successfully followed during the NAS and VA evaluation programs (see Procedure Chart).

THE PROCEDURE - USING THE STANFORD STORABLE CRUTCH AS AN EXAMPLE

To illustrate the validity of the evaluation procedure and provide an expanded description of the process, the Stanford Storable Crutch is cited as an example. As a brief introduction, this aid was designed and developed through the REC program at Stanford Children's Hospital in Palo Alto, CA, in response to the crutch user's need for portability and convenient storage. Constructed of an aluminum alloy, this axillary crutch employs telescoping mechanisms to reduce its length to 23 inches when not in use.

Step 1: Device Selection Process

- An Inquiry for Evaluative Services form was devised to collect preliminary information on potential evaluation items.
- Based on the responses from a survey of REC programs and the Inquiry form, the Storable Crutch was selected. Selection criteria considered were item function and application, target user population, state of market readiness, availability of evaluation support personnel and qualitative data.
- Stanford-REC initially contributed two third-generation prototype items for evaluation.
- A non-disclosure agreement between Stanford Children's Hospital and SwRI was signed to protect the developer's proprietary interests.
- FDA regulations were examined to determine test and evaluation compliance.

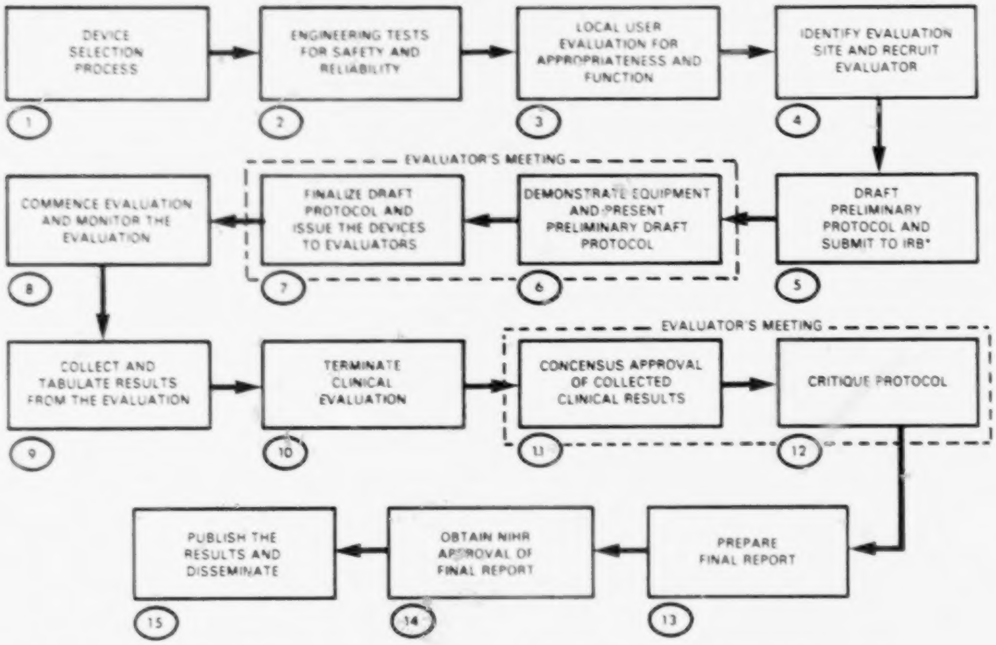
Step 2: Engineering Tests

- To assure crutch safety and reliability, a pre-check engineering analysis was performed.
- An engineering test and evaluation checklist encompassing criteria for human factors, strength, durability, and failure analysis (based on standards used by the U.S. Dept. of Defense) was developed.
- Drawings to identify components, materials, and assembly sequences were created to establish a common reference point for communication between the device developer and engineering test personnel.

Step 3: Local User Evaluation

- Two experienced crutch users were selected for participation in an informal evaluation conducted to determine potential functional problems.
- To encourage meaningful information feedback, an informal evaluation plan was implemented.
- The evaluators used the crutch regularly for approximately two weeks in accordance with the informal plan.
- Staff conducted a post-evaluation interview with the users.
- The users expressed high levels of satisfaction with the item, used both as a crutch and

SwRI-REC Evaluation Procedure


DEVICE SELECTION PROCESS
1
ENGINEERING TESTS FOR SAFETY AND RELIABILITY
2
LOCAL USER EVALUATION FOR APPROPRIATENESS AND FUNCTION
3
IDENTIFY EVALUATION SITE AND RECRUIT EVALUATOR
4
EVALUATOR'S MEETING
COMMENCE EVALUATION AND MONITOR THE EVALUATION
8
FINALIZE DRAFT PROTOCOL AND ISSUE THE DEVICES TO EVALUATORS
7
DEMONSTRATE EQUIPMENT AND PRESENT PRELIMINARY DRAFT PROTOCOL
6
DRAFT PRELIMINARY PROTOCOL AND SUBMIT TO IRB*
5
EVALUATOR'S MEETING
COLLECT AND TABULATE RESULTS FROM THE EVALUATION
9
TERMINATE CLINICAL EVALUATION
10
CONCENSUS APPROVAL OF COLLECTED CLINICAL RESULTS
11
CRITIQUE PROTOCOL
12
PUBLISH THE RESULTS AND DISSEMINATE
15
OBTAIN NIHR APPROVAL OF FINAL REPORT
14
PREPARE FINAL REPORT
13

a cane, because of its storability and durability. However, they identified two problems related to crutch stability that required minor design changes. Elements of these results were used to develop preliminary protocols for both clinical and long-term informal user evaluations.

Step 4: Identification of Evaluation Site

- Department of Physical Medicine and Rehabilitation (PM&R) staff at the University of Texas Health Science Center (UTHSC) in San Antonio were selected to oversee the clinical evaluation of the crutch because of the availability of appropriate users through their department. Subsequently, the REC and UTHSC staff enlisted the cooperation of the PM&R Dept. at the Audie Murphy Veterans Administration Hospital, also located in San Antonio.
- The fabrication of additional crutches through the Stanford-REC was requested. A minimum of two sets of crutches were required for clinical evaluation. An additional set would then be available for a long-term user evaluation outside of the clinical environment.

Step 5: Preliminary Protocol

- Based on the experience and results of the informal user evaluation, a preliminary protocol to structure and guide the clinical evaluation was drafted.
- The use of human subjects in any research program conducted through the UTHSC and the VA Hospital requires approval by their Institutional Review Boards. Evaluation plans, preliminary protocol, and all necessary paperwork were processed and approved.

Steps 6 and 7: Evaluator's Meeting

- REC staff conducted an initial evaluator's meeting with PM&R staffs to demonstrate the crutch and introduce preliminary clinical evaluation protocol.
- By concensus of REC and PM&R staffs, modifications to this draft protocol were minimal.

Step 8: Clinical Evaluation

- An additional set of crutches was received from the Stanford-REC and made available for evaluation.
- Candidate users were identified and screened by UTHSC and VA staffs.
- During the initial visit with the physician, therapist, and REC staff member, each user received instructions on crutch use and was properly fitted. The evaluation protocol was then reviewed.
- Using the finalized protocol as a guide, the evaluations were conducted for a two-week trial period with each user.
- Each user returned for a post-evaluation interview and completion of the evaluation protocol with at least one of the investigators.

Step 9: Tabulate Results

- Twelve adult crutch users completed the evaluation. User age, height, and weight ranged widely; disabilities, including both short and long term, varied.
- Generally, all results were related to the assessment of the acceptability of the crutch in its extended position as an ambulation aid, acceptability and performance of the collapsing/extending mechanism, and its stor-

ability in the collapsed position.

- A consultant statistician analyzed several of the questionnaire items, primarily those designed to permit ratings of crutch characteristics on a five-point scale.
- Subjective results were compiled, reviewed, and summarized by the UTHSC and REC staffs.

Step 10: Terminate Evaluation

- Within the constraints of a pre-determined schedule, clinical evaluation activities were terminated when a prescribed number of participants had completed both the device trial use period and the evaluation questionnaire, all post-evaluation interviews had been conducted, and resulting data had been collected and analyzed.

Steps 11 and 12: Evaluator's Meeting

- UTHSC and REC staffs reviewed all evaluation results, discussed problems associated with evaluation procedures and the protocol, and solicited comments and recommendations for improving the effectiveness and efficiency of the evaluation procedure.

Step 13: Prepare Report

- A complete account of evaluation activities related to the crutch was prepared.

Step 14: Approval of Report

- The program sponsor, NIHR, was informed of progress and problems throughout the course of the crutch evaluation. However, a final report of all activities was presented for the sponsor's review and approval.

Step 15: Disseminate Results

- Prospective manufacturers of the crutch were identified.
- An information package including a letter announcement offering the availability of this device for manufacture, a brief description of the features of the crutch, and pictures, was sent to these manufacturers.
- A technical support package, comprised of engineering drawings and evaluation information, and the prototype item, were made available to interested manufacturers.

Step 16: Negotiation and Contract with Manufacturer

(This additional step provides closure to this particular evaluation experience.)

- Based on interactions with interested companies, the developer selected a manufacturer to provide commercial availability of the Storable Crutch.
- SwRI-REC staff was available to discuss and clarify evaluation procedures and results related to the crutch.

CONCLUSIONS

If we are to meet our responsibility to develop and deliver appropriate, cost-effective technical aids to disabled people, then clearly, we must guide our developmental products through an objective, systematic evaluation process. The methods and procedure proposed by SwRI in this paper represent one example of a model that has demonstrated success. As the experience base with this model grows however, SwRI-REC staff anticipate making procedure revisions and refinements in an effort to create an effective, expeditious, coordinated, and results-oriented model.

Some organizations and programs have demonstrated equal success with other evaluation models. Shared experiences of success and failure with procedure and protocol development, communication flow, documentation methods, etc. may prove valuable.

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COMPUTERIZED ANALYSIS OF DEMOGRAPHIC DATA ON DISABILITY

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ABSTRACT

A computerized system was designed to allow demographic data collected by the National Center for Health Statistics in the 1979 National Health Interview Survey and Home Care Supplement to be reassessed according to variables of interest to the rehabilitation specialist. With this flexible system, populations may be defined in terms of categorizations based on functional limitation, medical etiology causing the limitation, and/or use of aids to supplement this limitation. Subsamples of the data can be organized according to any number of critical variables, including employment status, access to third party funding, and education level. The procedures developed in this project will provide access to information which can provide guidelines for decisions made at all levels, in the public and private sectors, concerning the needs of people with disabilities.

INTRODUCTION

The need for data which more accurately represent individuals with disabilities has been consistently expressed by rehabilitation specialists in both the private and public sectors. Valid and understandable data are essential if policy decisions regarding project funding and implementation are to reflect the needs of persons with disabilities. At present, reliable data on the disabled population are not readily available nor easily interpreted (1).

Analyzing data currently available is complicated by several factors (1). First, no classification scheme has been accepted in the field as the standard for operationally defining disabling conditions. Since different surveys have employed various operational definitions, comparisons among results are difficult. To further confound comparisons, data collection procedures vary among surveys. There are also practical difficulties of determining how individuals with multiple disabilities are classified and counted. Further, data are often collected from a subsample of the general population which does not include children or institutionalized individuals, therefore resulting in underestimates.

In this study, a computerized system was developed to allow data, previously collected on the nation's health (2), to be reanalyzed according to variables chosen to provide a greater understanding of the capabilities, limitations, and needs of people with disabilities. The data source selected for this analysis was the National Center for Health Statistics (NCHS). This governmental agency was established as a division of Health and Human Services to gather data on a regular basis on a variety of health related topics. In designing its nationwide surveys, NCHS employs experimentally sound procedures generating statistically valid and reliable data. Unfortunately, the reporting of the data often obscures details of importance to rehabilitation specialists. Frequently, data from several categories are combined into a single statistic thereby making the distribution of specific limitations among persons with certain disabling conditions unclear. The system designed in this study allows for manipulation of these data according to any variable of interest to the investigator.

In developing a demographic database, emphasis was placed on defining disability in terms of functional loss and limitation rather than medical etiology. With this focus, the impact of technological developments in the field of rehabilitation engineering can be better assessed. By defining disability in terms of functional categorizations rather than medical etiologies, the utility of a particular product for a given subset of the population can be determined. Since assistive devices are typically designed to overcome limitations in functional capabilities which may have resulted from any number of medical conditions, knowledge of medical diagnoses is useful only when it can be used to elucidate the nature of the functional loss.

MATERIALS AND METHODS

The Data

In an effort to develop a database from which data on disability could be drawn, raw data in computer tape format were purchased from NCHS. Public use data tapes from the 1979 National Health Interview Survey (HIS) and the 1979 Home Care Supplement were selected because they contain numerous questions concerning functional limitations, medical etiologies causing the limitations, use of aids, home care needs, and several other topics of interest. Further, although more current (1980) data are available, the 1979 Home Care Supplement is the most recent supplement related to disability for which a written, public report has been prepared. An important element in evaluating the accuracy and the value of these data involved a comparison of the results obtained when working with the raw data with those obtained from the published reports. Further, statisticians at NCHS have stated that the 1979 and the 1980 Home Care Supplements were conducted in such a way that the results from the two surveys may be combined at a later date without jeopardizing statistical validity. In fact, combining these two sources of data should increase statistical reliability.

Data from the HIS and the Home Care Supplement (2) are drawn from the same population, a nationwide sample of approximately 42,000 households, representing about 111,000 persons, in the civilian noninstitutionalized population. Only those individuals living during the interviewing period were included. The sample was designed in such a way that it is, theoretically, representative of the geographic, age, sex, and race distribution of the United States population.

The survey procedures entailed household interviews; therefore, all responses represent self-perceptions of the persons interviewed. In addition to collecting socio-economic data, NCHS asked a series of questions regarding the presence of specific medical conditions. A condition is defined by NCHS as a "departure from a state of physical or mental well-being." All conditions, except those considered impairments, are coded according to the Ninth Edition of the International Classification of Diseases (3). Impairments are chronic conditions that result in functional loss or limitation. If a condition could be considered an impairment, it was classified as such using special codes (called x-codes) developed by NCHS rather than with ICD codes.

When the survey was conducted, the questions on medical conditions were divided into six lists. The sample of households surveyed was randomly divided into sixths. Each sixth was asked one of the conditions lists. An exception to this procedure was implemented for those individuals who indicated that they were "limited in activity" because of a chronic condition. For these subjects, all six condition lists were employed. According to NCHS staff, this surveying strategy directly impacted on the statistical reliability of the population estimates. When estimating the number of people with a certain set of conditions, the critical variable to consider for error rate determination is whether the question of interest was asked of the entire sample or of one-sixth of the sample. The larger the sample surveyed the lower the error rate.

The Program

In developing software to manipulate these data, the sampling procedures employed by NCHS were considered. A goal of the project was to develop a system which would allow a high degree of flexibility in defining the population of interest, without sacrificing the statistical reliability of the population estimates. The size and structure of the data files complicated the development of the program. The public use files for the 1979 Health Interview Survey and Home Care Supplement reside on eight reels of magnetic tape. The set consists of six files: separate data files for households, persons, homecare, medical conditions, hospital visits, and doctor visits. Use of the files in this format is inconvenient, especially when information from multiple files is desired. In addition, identical data about persons appear on four of the files, and, therefore, during any given retrieval, approximately 60% of the data processed is redundant. In order to improve processing efficiency and to provide convenient statistical analysis capability, the public use files were transformed into a simple hierarchical file containing no redundant data. This step reduced the total data by over 80%.

A two step software system was then developed to support the selection and tabulation of the reduced data. The first step in the software system involves a generalized retrieval program which is easily tailored to each unique analysis project. The retrieval program selects the subset of interest from the hierarchical file and produces a simple flat file suitable for tabulation and other statistical processing. This step further reduces the data to include only the information needed for a specific analysis. The second step of the system consists of two general purpose tabulation programs. One of the tabulation programs requires more effort to tailor it to a specific project but also offers extensive flexibility and low computer costs. The other tabulation program offers easier initial setup but higher computer costs. Both programs compute relative standard error for each statistic computed. Error rates are estimated by using equations derived by NCHS which take into account their sampling procedures.

The entire software system is designed to run on an IBM mainframe under the MVS operating system. The software is written in PL1[1] and SAS[2]. In addition to these languages, the host system must also support a sort utility in order to run the program. All software and documentation developed through this project are noncopyrighted and in the public domain.

RESULTS

Trial runs of the system have demonstrated its flexibility. Populations may be defined in terms of categorizations based on functional limitation, medical etiology causing the limitation, and/or use of aids to supplement this limitation. Further, it has been shown that subsamples can be tabulated according to any pre-selected NCHS variables. In addition, the accuracy of the system was evaluated by comparing the estimates obtained from manipulating the raw data with data presented by NCHS(2). Examination of the data indicated that the estimates generated by the program are consistent with the published data.

An assessment of the reliability of the estimates suggested that constraints should be built into the system. To maintain flexibility but still ensure the statistical reliability of the estimates, the entire sample can initially be screened to include only those individuals who indicated that they were "limited in activity" because of a chronic condition. Only these individuals were asked questions from all six condition lists, and, therefore, estimates based on this subsample will be subject to less statistical variance. This constraint allows for an estimation of the prevalence of chronic conditions within the population of individuals who are limited in activity.

To estimate the prevalence of either acute or chronic conditions for the general U.S. popula-

[1] PL1 is a product of the IBM Corporation

[2] SAS is the registered trademark of SAS Institute, Inc., Cary, NC

tion, the screening feature may be bypassed. However, to derive these estimates, the condition of interest must be defined in terms of variables from only one condition list. Therefore, one could determine the prevalence of a specific condition but could not necessarily determine what other conditions individuals in this population may have. Further, needs for home care for this population could not be accurately assessed.

DISCUSSION

The procedures developed in this project allow for an estimation of fairly well defined segments of the disabled population. The system is, of course, limited to the extent that all inquiries must conform to the format of the survey employed by NCHS. However, an examination of the questionnaire utilized in the survey reveals that a wide range of topics were covered.

In the context of rehabilitation engineering, perhaps the most direct application of the database is the estimation of the numbers of individuals who might benefit from the use of a specific technology. It is the population of chronically affected individuals, perceiving themselves as having an activity limitation, who probably are most likely to be the users of aids and devices.

The database also has much broader applications. Within this data set is a wealth of information on the disabled population. By creatively extracting and crosstabulating variables, one can begin to develop demographic profiles of this diverse group of people. By utilizing data previously collected, information can now be organized in terms of any number of critical variables, such as employment status, access to third party funding, education level, and so forth. Access to these data can provide guidelines for decisions made at all levels, in the public and private sectors, concerning the needs of people with disabilities.

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7.5

DISSEMINATION ISSUES FOR MEDICAL ROBOTS

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ABSTRACT

Numerous factors will determine the successful diffusion of robot technology into the health and human service market. This paper will present an overview of some of the issues that will need to be considered as robotic technology is incorporated into health care services.

DISCUSSION

When will robots reach the health and human service market and **to what extent** will they be disseminated?

Overall, robots appear to be in a development stage similar to that of business computers in the early sixties. Like the computer pioneers, the robot marketers' success will depend on a number of factors both within and outside their control. The subtleties of positioning and introducing programs will be critical. The adaptability of current operations will be key, and, of course, acceptance by the ultimate human controller.[1]

The realization of health care robots will depend on financial and human resource commitment to the research and development of health and human service applications and the extent of investment in the design and manufacturing of systems to meet those applications.

- There is no *demand* for robots in the traditional sense in the health and human service market. We will need **research** that will predict the extent of potential cost savings, increased work efficiency, increased quality of care and better work conditions. In particular, unsafe, high risk, or odious tasks could be performed by robots. The creation of a viable market with a wide spectrum of niches will depend on numerous factors. As mentioned, application areas must be clearly identified and the appropriate robot design must be created for these demands. Market niches will be carved out by those systems that can provide the best *fit*. *On the microeconomic scale, the advantage has been and will continue to be, with the company that can "build a better mouse trap" and offer it to the consumer at a "better price".*[11] **Demand analysis** will be needed to identify which factors are most influential in determining the categories/features of robot technology that people will need and are willing to purchase.

Market

- The potential health care **market** alone is significant. The Office of Technology Assessment estimates that there are between 15 and 25 million people with disabilities in the United States[14] who, to varying extents, may have need for a robotic assistant. There are nearly 7000 hospitals[2] and approximately 20,000 nursing homes and other longterm care facilities in the United States. Approximately 50% of hospital operating costs are in labor-related expenditures.[2]

In the current 'cost-containment' health care environment, robotic devices which increase productivity and reduce labor costs are more likely to be accepted.

Costs

- The benefits of flexible automation have been realized in numerous industrial settings. The potential for reducing health care **costs** through augmentation of human resources is a strong incentive for further exploring development of robotic tools in this area. There have been chronic difficulties with high employee turnover rate in longterm care institutions. There are also significant problems with finding qualified, reliable personnel at the *hands-on* care level. This problem will be compounded as the needs of a rapidly rising older and disabled population put increasing demands on these care facilities.

- The market will support devices with a range of **prices**. *The cost of robots varies, of course, as a function of the robot's complexity and versatility...*[10] From the **individual** purchaser's perspective: *There are four conditions which must be met by a Robotic Aid before it will be an* **economically feasible** *partial substitute for human caretaking:*
  - *(1) The* **manipulative capability** *of the aid must be sufficiently good to make it an attractive alternative to some classes of human assistance.*
  - *(2) It must be* **reliable** *enough that the user and his/her attendants will be encouraged to use it.*
  - *(3) Savings (and earnings) derived from using the robotic aid must* **pay for itself**.
  - *(4) The system must be psychologically and* **socially acceptable** *to the disabled user, the users' family and associates.*[12]

- The market can support a range of **costs** from the **institutional** purchaser's perspective. Volume purchases are also possible because of the growing number of institutional chains which already contract for services or supplies on a collective basis. Government contracts, such as from the Veterans Administration, with its large numbers of hospitals and nursing homes, are also a major source of volume sales. Continued research on third party reimbursement of evolving technologies that might be classified under DME (durable medical equipment) will provide useful information to developers and policy decision-makers. For instance, perhaps third party funding may soon reimburse for expenditures on environmental controls (these are not presently covered under Medicare reimbursement procedures, because they are generally not deemed *medically necessary*); it would be important to build this capability into a robotic aid. Demonstration of *medical necessity* (which requires physician prescription and physician verification) will be essential for a robotic aid to assist disabled persons since over two-thirds of our disabled citizens depend on third parties such as Medicare and Medicaid for medical care and assistive device funding.[3]

Cost-benefit analyses are useful only when both costs and benefits can be estimated fairly well, and when an appropriate technology exists to absorb the costs and produce the benefits... The costs averted by applying a procedure must be balanced against the risk averted by not doing it.[7] Cost studies with health and human service robots must include **risk** as well as costs and benefits .

Methodologies

- Over one third of all products researched and developed for rehabilitation never reach the market. There is a serious need for **methodologies** to be developed which will facilitate a rapid and smooth research, development, evaluation, and dissemination cycle for the proposed product, whether its development is privately or publicly funded.[13] It is difficult to separate design and dissemination issues. For our purposes, they may be viewed as two sides of the same coin. Interrelated is the sense of useful products reaching appropriate markets in viable numbers for commercial success.

Diffusion Strategies

- Developing reliable devices that can serve in a variety of settings will determine the marketing strategies that will need to be adopted. The primary barrier to the diffusion of robots into health and human services will to overcome the fear and/or reluctance of the proposed end-users. *Robotics— indeed all technologies that result in benefits to the consumer- have critical consequences and challenges for marketing. Specifically, marketers must identify and communicate those benefits to consumers.*[10]

Government's Role

- The time frame for robots reaching the health and human service market will also depend on the role **government** will play in fostering the industry. Some of Japan's success with robotic applications has been due to the Japanese government's interest in and encouragement of the robotics industry. It helped establish the Japan Robot Leasing Co. which provides incentives, such as, short-term robot leases and an additional 13% depreciation on purchased robots, to potential robot users (manufacturers, primarily).[17] Government interest could help impact strongly in the research, development and evaluation of robotics.

Professional Acceptance

- Innovation diffusion takes place *in the context of a social system.*[16] Medical professionals have traditionally been receptive to incorporating new technologies into their diagnostic and therapeutic regimes. Patients often expect certain examinations and therapies which include recent technological breakthroughs. It is likely that a robot will face little resistance in health care settings if it is carefully introduced. Health care professionals respect the opinion of their peers; therefore, it will be important that individuals experienced in health care delivery play a significant role in introducing robots into health care settings.
- **Training** will play a critical role in acceptance and utility.[4] *As the level of awareness of robots and their usefulness and applications increases, it is imperative that education and training activities provide individuals with the necessary understanding of robots and various automation systems, as well as provide involvement in the process.*[9]

Facilitating employee acceptance of these robots into the work place is a major task facing manager of human resources.[11] Acceptance is a diffusion and marketing question. The question of employment and retraining (inherent in a technological evolution that requires changing patterns of required skills and new approaches to knowledge acquisition) can cause employee and management uncertainty. James A. Baker stressed the importance of a positive approach to this phase of the diffusion process. Commenting on GE's reasonable success with the interaction between hourly employees and robots, he stated that GE had followed the Japanese management technique of stressing the *positive* effects of robots. He also discussed employees early involvement in the dissemination plans. He stated that they:

> *were taking pains to get them* [employees] *and their representatives on board early in our planning to introduce the devices into our plants. We've set them up on display in plant cafeterias so our workers could become familiar with them before installation. Their response has been positive for a number of reasons. First of all, they feel excited and proud to be working with -and supervising, in most cases- some of the most advanced technology in industry. Kids whose fathers spray paint refrigerators by hand don't tend to be the hit of the schoolyard. Kids whose fathers train and control robots do.*[5]

Communication Channels

A channel is the means by which a product (message) gets from a source to a receiver.[16]

- We presently have a communication gap between health and human service professionals and robot manufacturers. That information exchange gap can be addressed with two types of communication channels: mass media and interpersonal. Communication channels are particularly important during the first stages in the innovation adoption process.

Mass Media Channels

- The mass media plays a pivotal role in determining public attitudes toward robots. *Mass Media channels: (1)reach a large audience rapidly, (2)create knowledge and spread information, and (3)lead to changes in weakly held attitudes.*[16] *There is little doubt among people in the industry that R2D2 and C3PO, the friendly androids of the Star Wars trilogy, have up to a point helped to provide a more open and responsive environment to market robots in... In the long run, however, there is little doubt that the Star Wars imagery will be good for the industry. As one source points out, the teenage audience tht forms the greatest box office support for films like Star Wars will be the ones in a position to purchase home robotic technology when it comes into existence in a few decades.*[1]
- Continued media coverage that presents robot technology in a positive light can significantly impact on the diffusion rate. The tradional monsteresque representation is only recently giving way to the friendly, helpful R2D2 image of Star Wars. At least one television show has portrayed a *robot nurse* as having undesirable *robotality* (personality characteristics) more commonly associated with the *very human* head nurse in **One Flew Over the Cuckoo's Nest**. This has been counterbalanced by positive presentations of current robotics research by news and educational programs.
- Professional robotics associations such as the Robotics Society of America, the Robotic Industries Association, and Robotics International can serve as information conduits to the general public, to health care professionals, and to robot enthusiasts. They can help promote positive uses of robotic technology and reduce societal hesitancy toward dissemination of these important innovations.

To restate the obvious: successful dissemination into the health and human service arena will not depend simply on engineering expertise. It will require a coordinated effort of many disciplines and industries in order to identify and implement the latest robotic technology for use in a wide spectrum of settings.

Interpersonal channels:

(1)provide a two-way exchange of information. One individual can secure clarification of additional information about innovation form another individual. This characteristic of interpersonal networks sometimes allows them to overcome the social-psychological barriers of selective exposure, perception and retention. (2)Persuade an individual to form or to change a strongly held attitude. This role of interpersonal channels is especially important in persuading an individual to adopt an innovation.[16]

- The strength of the **professional** interpersonal channels is an example which exists in the medical industry. The whole concept of a *profession* is based on maintaining *professional norms* which are reinforced through the interpersonal channels within a facility and at medical conferences. The success of a robotic physician assistant may depend on its successful introduction at professional medical conferences. Continuing education courses as well as medical school training will be important avenues for informing physicians about the potential of robots; and at the same time, provide robot developers opportunity to gain insight on physician needs and attitudes toward robotic technology. This will apply to other medically related professional and allied health fields. For example, nursing and social work each have their own professional organizations and educational milieu into which robots will need to be introduced.

CONCLUSION

Robots are just beginning to be considered serious tools for service roles to humans. The extent to which they will be disseminated, and how soon, will depend on multiple factors. While there is a strong potential market, there is no market demand at the moment. Overcoming fear of and hesitation toward the unfamiliar may be two of the more significant challenges to future health and human service robot manufacturers and marketers.

ACKNOWLEDGEMENTS

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THERAPEUTIC EFFECTS OF ELECTRICAL STIMULATION IN SPASTIC CEREBRAL PALSY

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ABSTRACT

A home program of electrical stimulation was instituted to determine the therapeutic effects of electrical stimulation in 12 subjects with spastic triceps surae. Stimulation was applied for 30 minutes daily over a period of up to 18 months. As determined by gait analysis, no improvement in the degree of dorsiflexion during swing was noted when the stimulators were not used.

INTRODUCTION

The therapeutic effects of electrical stimulation (ES) in decreasing involuntary action and facilitating volitional movement have been commented on by many researchers. Such effects include: prolonged post-stimulation increase in voluntary power in foot dorsiflexion (Vodovnik and Rebersek, 1973; Carnstam et al, 1977); relaxation of spastic muscles following stimulation of their antagonists (Lee et al, 1950; Levine et al, 1952; Dimitrijevic et al, 1968); and improvement in the degree of postural sway (Van Griethuysen et al, 1982). Adults with hemiplegia resulting from stroke have been the major focus of research into short and long-term restoration of motor function through the use of ES. Evaluation of the therapeutic effects of ES has been through: the use of subjective clinical measures (Merletti et al, 1979; Alfieri, 1982); the measurement of poststimulation isometric torque (Teng et al, 1976; Carnstam et al, 1977); changes in the cyclical patterns of muscle activity following stimulation (Dimitrijevic et al, 1968); and kinetic and kinematic measurements (Malezic et al, 1981). While few of the studies referenced above have demonstrated meaningful statistical results, the many publications dealing with therapeutic results of ES in hemiplegia attest to the interest in this subject.

The influence of ES in stroke hemiplegia suggests that this modality may be effective in positively influencing motor function in children with spastic cerebral palsy. Gracanin (1978) reported on the use of peroneal nerve stimulators in 281 children with cerebral palsy. Stimulation was applied daily for several hours at a time. Of 190 children, improvement in gait and posture occurred in 150 children whilst 78 demonstrated long-term improvement. Details on how improvement was measured were not given. Riso et al (1980) reported on 6 children with cerebral palsy, 4 of whom had undergone daily stimulation to elicit dorsiflexion in the swing phase of gait for a year. Footswitch patterns were used to determine changes in foot-floor contact patterns and timing. No therapeutic effects of stimulation could be demonstrated. In a subsequent report, Riso and Makley (1981) described the attainment of therapeutic "carry-over" in one hemiplegic and one diplegic subject out of a total of seven children.

Based upon the encouraging results reported by Gracanin (1978) and Riso and Makley (1981), we undertook a study to determine the potential of ES to improve equinus gait in subjects with cerebral palsy.

METHOD

Twelve subjects underwent stimulation of the peroneal nerve for 30 minutes daily. Details regarding each subject are listed in Table 1.

Table 1 - Subject Details

	(a)		(b)		(c)
Subj.	Diag.	Sex	Age	Length of Stim.(wks)	Stimulator & side (d)
1	CP-D	M	12y 7m	32	P/B
2	CP-D	F	14y 9m	71	W/B
3	CP-D	F	9y 11m	32	W/B
4	CP-H	M	19y 9m	34	W/R
5	CP-D	M	47y 8m	23	W/B
6	HI-H	F	17y 5m	61	W/R
7	CP-D	F	8y 5m	13	P/B
8	CP-H	M	6y 4m	26	P/R
9	CP-H	M	13y 11m	7	W/R
10	CP-D	M	4y 5m	9	P/B
11	CP-H	F	13y 5m	45	W/L
12	HI-H	M	10y 8m	56	W/L

(a)
CP-cerebral palsy
HI-head injury
D -diplegic
H -hemiplegic

(b)
Age at start of stimul.

(c)
W-walking
P-seated

(d)
B-bilateral
R-right
L-left

In 6 subjects, ES was applied to the foot dorsiflexors bilaterally in a reciprocal manner with the subject seated. Seven subjects were provided with dual-channel stimulators to elicit dorsiflexion during the swing phase of gait. A stimulus frequency of 40Hz and pulse widths of 250s were used. The stimulators are described in Naumann et al, 1984. Subjects or their parents were instructed in the use of the stimulators prior to taking them home.

Because of initial low compliance (a total of 19 subjects entered the study), home visits by staff were instituted. At first, each subject was visited three times a week. As subjects or their parents became more adept at using the stimulators, visits were reduced to one per week.

Each subject was assessed in the gait laboratory prior to the commencement of stimulation and then at approximately monthly intervals throughout the stimulation period. Assessments were performed following 24 hours during which stimulators were not used. Data collected included foot-floor contact patterns, the electromyographic activities of quadriceps, hamstrings, gastrocnemius and tibialis anterior muscles, and the time-histories of joint marker trajectories. Assessments were performed with the subjects barefoot. At least 3 footsteps per side were analyzed on each occasion. Positional data were reduced to values occurring at the end of stance, the end of swing, the maximum and minimum. Sagittal angles for the hip, knee and ankle and the angle between the floor and the foot were determined. The ratios of the direct distance between the start and end points of each marker to the actual path traced out by the marker in the transverse plane were determined as a measure of medial-lateral movement. The percentage of stance and the stance-to-swing ratio of the left to right hand sides were calculated from an average of 12 footsteps per occasion.

The 7 subjects who used ES during walking also had their gait on occasion evaluated with and without stimulation.

RESULTS

The accuracy in determining joint angles was ± 2.4°. Thus changes of less than 5° were set to zero for analysis purposes. The change in ankle dorsiflexion across all 7 subjects with and without stimulation at heel contact showed an average increase of 7.2° ($p<.025$). The average increase towards dorsiflexion in the angle between the floor and the foot was 5.2° ($p<.01$). The maximum of this angle increased by a mean of 6.23° ($p<.01$). No significant changes were found in hip and knee angles across these 7 subjects. Other positive changes that were significant in certain individuals but not across all subjects were internal/external rotation of the foot and its degree of valgus/varus.

Each parameter derived from the monthly gait assessments performed without stimulation was fitted to a linear curve using the least squares method and the F-value and its significance was then calculated. Figure 1 is an example of the fitted curves. Stance times as a percentage of the total gait cycle are shown versus time. Changes between the start and end of the stimulation period that were less than 5°, even when the F-value was significant, were again set to zero. For analysis purposes, each stimulated side was considered independent so that 18 sides were considered in total.

Having determined significant changes in parameters for each side, changes in each parameter were tested for significance across all sides. Only the ratios of direct-to-actual distance travelled by markers in the transverse plane showed significance at the .025 level. An analysis of variance revealed no significant relationships between parameters or groups of parameters at the .05 level or less. Nor did the type of stimulation used or the age of the subjects show any influence.

DISCUSSION

Although not significant, there was a trend to increased hip, knee flexion and/or ankle dorsiflexion during and at the end of the swing phase. This indicates that the decrease seen in medial-lateral movement may be the result of ES facilitating an increased withdrawal reflex of the lower limb. Even though the number of subjects was small from the statistical point of view, the results confirmed our subjective impression that ES did not appreciably improve

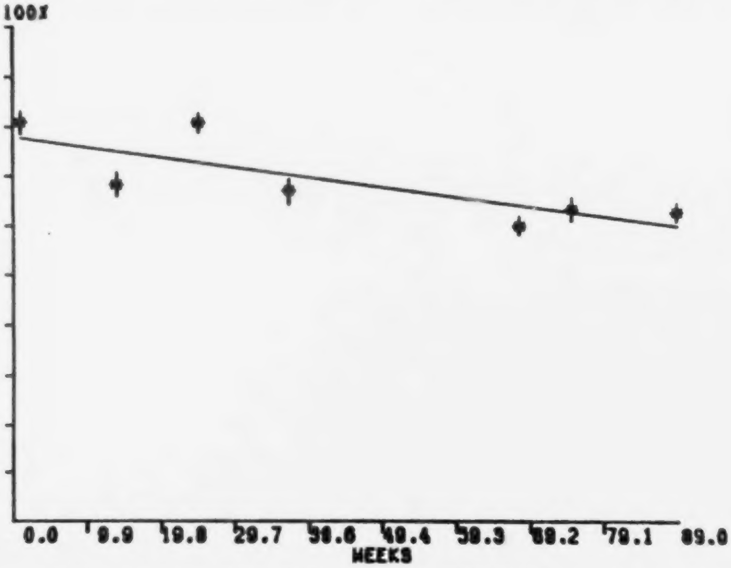

Figure 1. % stance versus time on right hand side for subject #2.

the quality of gait in these subjects. No attempt was made at obtaining a control group since it was improbable that a comparable group of people could be found. Thus, each subject acted as his/her own control with changes in calculated parameters being used to determine the effects of ES. This had the disadvantage that, since the effects of spasticity such as the degree of joint contractures may increase with time, no significance could be given to a lack of change in the stimulated subjects.

Both short-term and long-term benefits can be expected from ES. Short-term benefits may result from inhibition of antagonistic muscles (in this case, triceps surae). In our subjects, we noted that if the stimulators were turned off during walking, the effects of stimulation would disappear within three footsteps. Long-term benefits could result from: (i) increasing the strength of stimulated muscles which may otherwise be inhibited by their spastic antagonists. We noted an increase in the strength of the dorsiflexors as measured clinically and an increase in the degree of volitional dorsiflexion in isolation of other movements over the stimulation period. However, these increases in power and range did not carry over into gait suggesting that their influence was overridden by the established reflex patterns; (ii) reorganization of reflex motor activity through the inflow of sensory signals and/or the inflow of normalizing patterns of movement to the central nervous system which, if repeated a sufficient number of times, may result in the establishment of a more normal "engram" (Gracanin, 1977, 1978). The decrease in medial-lateral movement evident in our subjects may be the result of such a reorganization. In addition, in 2 of our subjects, we noted the appearance of tibialis anterior activity at the end of the swing phase following several months of stimulation. Again these changes did not appreciably improve the quality of gait in these subjects. It could be argued that, since hundreds of thousands of repetitions are necessary for the establishment of a pattern of movement (Kottke, 1980), patterning through ES for 30 minutes daily is insufficient to achieve this. Also, it may be necessary to control more than the ankle joints through the use of multichannel stimulators for this treatment to be more effective. These hypotheses cannot be tested, however, without first solving current problems associated with the use of surface electrodes. These problems also affect the use of ES as an orthotic aid and are: the time needed to correctly locate the electrodes over motor points; lack of cosmesis especially in summer time when children are active outdoors; skin-breakdown if the electrodes are in position for too long. Given the problem of skin-breakdown and the time needed to apply electrodes, subjects found it difficult to devote more than 1 hour per day to ES. Ten of the subjects had to be constantly encouraged to continue using the stimulators because of lack of improvement in their gait due to ES. Only 2 subjects requested they be allowed to continue ES following the study period because they felt it to be of benefit.

CONCLUSION

The application of ES for 30 minutes daily over a period of up to 18 months in 10 subjects with spastic cerebral palsy and 2 with head injuries did not improve the degree of dorsiflexion attained during swing nor lead to the attainment of heel strike when the stimulator was not used.

ACKNOWLEDGEMENT

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8.2

KINEMATICS OF PARAPLEGIC GAIT PRODUCED BY ELECTRICAL STIMULATION

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ABSTRACT

The effect of electrical stimulation on the kinematics of a walker-supported paraplegic gait was studied both with the basic stimulation pattern including hip flexors, knee extensors and dorsiflexors; and with an augmented pattern where hip extensors and abductors and plantar flexors were added. The subject was better able to transfer his weight with the augmented pattern. The results showed a 40% increase in the stride length, 2.5 times increase in speed of walking, decrease in time of double support and less transfer of weight onto the walker.

INTRODUCTION

Paraplegic walking in a swing-to or swing-through gait using electrical stimulation (2,6) has little advantage over walking with long-leg braces. In both cases the knees and sometimes the hips are locked in extension - by electrical stimulation of the quadriceps and gluteal muscles in one case and by braces in the other. Locking the knees and hips with electrical stimulation places a further energy demand on the patient in addition to that required for ambulation. Long-leg braces, by themselves, are usually abandoned by paraplegic individuals with levels of injury above T11 due to the large amounts of energy required for ambulation (7).

Providing paraplegic patients with reciprocal gait which is energy efficient and cosmetic in appearance is a desirable alternative. A reciprocal gait has been produced in paraplegic patients using both surface (1,3) and intramuscular electrodes (4) for activation of muscles. With surface stimulation knee extension and hip flexion were the two main functions provided. The use of intramuscular electrodes provided a more precise control over individual muscles and made hip extensors and abductors and plantar flexors readily accesible. The purpose of this paper is to compare the kinematics of the reciprocal gait produced by electrical stimulation of hip flexors, extensors and abductors, knee extensors, and ankle plantar and dorsiflexors; to the gait produced by stimulation of hip flexors, knee extensors and ankle dorsiflexors alone.

METHODS

A paraplegic subject, weighing 73kg and 1.75m tall, with complete absence of motor and sensory function below the T8/9 level was implanted bilaterally with intramuscular electrodes. The semimembranosus and gluteus maximus were implanted for hip extension; the sartorius, gracilis, and tensor fasciae latae for hip flexion, the gluteus medius for hip abduction; the quadriceps, excluding the rectus femoris, for knee extension; the tibialis anterior and peroneus longus for dorsiflexion; and the soleus for plantar flexion. The right gluteus medius was much weaker than the left.

A stimulation pattern for walking was determined by trial and error using a computer, on the basis of muscle activities during walking of normals. A videotape recorder was used to study the qualitative appearance of the walking. The stimulation pattern was adjusted until the most naturally appearing gait was attained. This pattern of stimulation was transferred into a portable 32-channel microprocessor-controlled stimulator (8). The stimuli were biphasic, charge-balanced, constant-current pulses of 20 mA in amplitude, modulated from 0 to 150 microseconds in pulse width, at a frequency of 22.5 Hz.

The subject was evaluated twice; first, using the basic stimulation pattern for hip flexors, knee extensors and ankle dorsiflexors only. Second, hip extensors and abductors and plantar flexors were added to the basic pattern. The subject was evaluated in the Gait Laboratory on a 5m instrumented walkway covered with conductive rubber; and conductive tapes were placed on the subject's shoes over the heel and the medial and lateral aspects of the forefoot to determine foot-floor placement simultaneously with joint angles (5). The knees and ankles were instrumented with goniometers to measure angles in the sagittal plane. Speed of walking, time of double support and stride length were also measured.

A mock-up walkway was constructed with force plates to measure vertical forces (Fz1 and Fz2) exerted at the foot-floor contact; and the amount of body weight transferred to the rolling walker, used for balance and support during walking. Data collection was synchronized with the preprogrammed pattern of muscle stimulation through a switch which initiated the swing phase and allowed the subject control over progression through the pattern.

RESULTS

Qualitatively, the gait produced by inclusion of hip extensors and abductors and plantar flexors in the stimulation pattern (Fig. 1) appeared more normal and was less energy demanding of the subject. He felt more stable and was better able to transfer his weight from one leg to the other. This was shown in better repeatability in the foot-floor contact, increase in the speed of walking and stride length, and decrease in the

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time of double support. In both cases the stance phase was initiated with heel contact followed by a foot-flat. The swing phase was initiated from either lateral plus medial or medial contact. Irregularities were observed in the foot-floor contact during stance phase when using the basic pattern of stimulation only. The speed of walking increased from .16 m/sec with the basic pattern to .4 m/sec with the augmented pattern. At the same time the stride length increased from an average of .7m to 1m. Commensurate with the increase in speed the time spent in double support from right heel strike to left toe-off decreased from 1.24 sec to .58 sec; and double support from left heel strike to right toe-off decreased from 1.54 sec to .74 sec.

For both patterns the swing phase was initiated by the subject after he actively transferred about 70% of his weight to the legs and the rest to the walker. The duration of the swing phase was fixed by the preprogrammed pattern of stimulation. During this phase the knee went through maximum flexion of 65 degrees. It took an average of .6 sec to achieve maximum hip flexion from the time of hip flexor activation; and another .6 sec to achieve full knee extension from the time of quadriceps activation. Weight distribution before initiation of the swing phase and maximum knee flexion during swing were about equal with both stimulation patterns. At the completion of the swing phase, the knee was fully extended or slightly hyperextended. The contralateral knee was fully extended with the ankle dorsiflexed up to 10 degrees.

The ankle was plantar flexed 10 degrees to 15 degrees less with the basic pattern before the initiation of swing; and 10 degrees to 15 degrees more following the swing phase. With the augmented stimulation pattern, the weight distribution as determined by vertical force measurements (Fig. 1) indicated that up to 100% of body weight was borne on the legs at completion of the swing phase, before forward weight transfer. With the basic pattern of stimulation, a maximum of 80% of the body weight was transferred to the legs. The weight transfer with the augmented pattern was initiated while the knee was still flexed at about 20 degrees. This did not occur with the basic pattern.

Similar joint angles and foot-floor pressure distribution were observed during the second swing phase and double support. The whole cycle was then repeated.

DISCUSSION

These results show the importance of inclusion of hip extensors and abductors and plantar flexors in the pattern used for electrically produced walking in paraplegics. Use of these muscles has increased the subject's ability to transfer weight from one leg to the other enabling him to take longer strides, to walk faster, and to spend less time in double support. Evaluation and adjustment of the gait by means of repeated video-tape observations has its limitations; and this is obvious from the asymmetric stimulation pattern that resulted with this method. The double support data indicated that due to the asymmetry in the pattern, the subject was better able to transfer weight from left to right foot, than the other way around. The modifications that may improve the weight transfer in this case are the addition of strong right hip abductors during left swing phase and the activation of right hip extensors during double support.

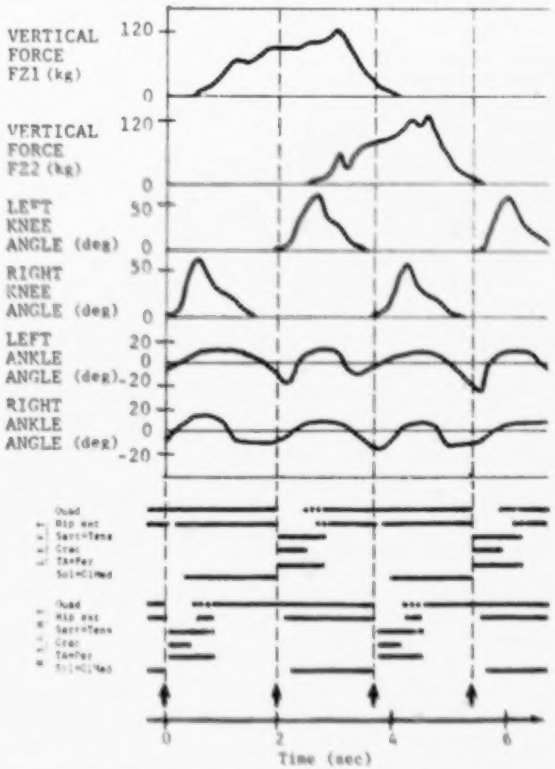

Figure 1 Vertical Force, Knee and Ankle Angles with Augmented Stimulation Pattern (arrows indicate subject's commands to initiate each step and dotted lines indicate pulse width modulated between 0 and 150 microseconds)

Some of the limitations in increasing the speed of walking are the amount of time that it takes to achieve a desired motion such as hip flexion and knee extension after the muscles have been activated. Higher stimulation frequency can improve the response of the muscle but it may do so at the expense of early fatigue. Further, better hip extension, abduction and push-off will increase the forward transfer of weight. Currently only the soleus is used for plantar flexion which in normal individuals accounts only for about 50% of the total plantar flexion force. Increasing the strength of the muscles used through electrical stimulation exercise may also further improve the walking.

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8.3

CORRELATION OF CLINICAL PROFILE WITH GAIT PARAMETERS IN THE SCI POPULATION: A BASIS FOR FES PATIENT SELECTION?

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ABSTRACT

Improvement in FES gait assist success requires the ability to select appropriate candidates as well as to incorporate the patients' residual function within the stimulation plan. Correlation of clinical features of the SCI patient with gait effectiveness has not been documented. Accordingly, six clinical characteristics were evaluated in a group of 19 incomplete SCI patients and findings were related to velocities and joint motions recorded during gait. Spasticity, selective muscle test strength, upright motor control and patterned muscle test strength were found to be associated with velocity, and swing hip and knee flexion obtained. Gait performance was unaffected by the proprioceptive or range of motion characteristics of this group.

INTRODUCTION

The existing SCI population in the United States has been estimated to number from 150,000 to 500,000 persons. Seven to ten thousand new injuries occur each year (1). Despite treatment by bracing and therapy, ambulation is achieved in only 25% of the population (2), reflecting a need for a better solution to this problem. Current research in the use of functional electrical stimulation (FES) to activate centrally denervated muscles has suggested that FES may be a viable modality for restoration of ambulation in selected SCI patients. While preliminary FES efforts have resulted in slow gaits, improvements are expected in the future with greater use of implanted systems and improved stimulation plans.

Kralj (3) and Stanic (4) have suggested that FES effectiveness will be enhanced by inclusion of the patients' limitations as well as arsenal of preserved movements into the stimulation sequence. Most commonly, attempts to identify residual capabilities have utilized observational gait analysis or clinical test results. The main limitation in using observational gait analysis to define residual function is its inability to differentiate weak or absent agonist action from excessive antagonist action or limited mobility. Hence, clinical testing also is indicated but correlation of specific clinical features with walking ability has not been well documented for this population. The purpose of this study was to relate the clinical features of patients with incomplete SCI lesions to gait effectiveness achieved. Knowledge of the effects of clinical characteristics on gait performance should enable a more informed selection of suitable FES candidates as well as stimulation regimes.

MATERIALS AND METHODS

Nineteen patients with incomplete SCI lesions and inadequate swing flexion were identified from Rancho Los Amigos Medical Center outpatient SCI clinics to participate in this study. Level of lesion varied from C4 to T12. Functional ambulation classes spanned physiological to unlimited community ambulation with gait velocities ranging from 3 to 60% of normal.

Six clinical test procedures were performed to identify the clinical profile of each patient. Patient response was monitored by dynamic EMG, motion and stride analysis. Test findings were related to patient gait characteristics.

The residual motor power that could be activated by the patient was evaluated by means of standard manual muscle tests to identify selective control, patterned strength tests to measure mass flexion and extension capabilities and upright motor control (UMC) tests of the hip, knee and ankle to detect primitive stepping and stance mechanisms. Performance on both selective and patterned strength tests was graded using conventional criteria. These grades were later converted to percent of normal torque equivalents developed by Beasley (5).

Range of motion tests assessed passive mobility. Quick stretch tests of the hip, knee and ankle antagonists identified the degree of obstructive spasticity present. Proprioception was checked at each joint to detect areas of impaired or absent position sense.

Muscle activity during strength testing, UMC testing and gait was recorded from hip, knee and ankle musculature by means of dynamic intramuscular electromyography. Lower limb motion was evaluated in the upright motor control tests and gait by means of a 2-dimensional motion analysis system utilizing reflective joint markers to define pelvic, thigh, knee and ankle sagittal plane motion. Compression closing footswitches were used to record stride characteristics of the patients' gait. Footswitch data were telemetered to a VA-Rancho stride analyzer for processing and displayed on visicorder paper to serve as a timing reference for muscle activity.

Statistical Analysis

Two sample t-tests were used to assess the effects of differences in discrete clinical attributes on gait velocity and motion. Where clinical data was ordinal, Pearson's correlation tests were utilized. All statistical tests were performed with an RS1 software package running on a PDP 11/34 computer.

RESULTS

Clinical features which were found to influence velocity and swing hip and knee flexion included spasticity, selective muscle strength, upright motor control and patterned muscle test strength. Gait performance was unaffected by proprioceptive or range of motion characteristics of this group.

Spasticity

The presence of either moderate or severe extensor spasticity of hip, knee or ankle was associated with decreased walking velocity. Persons without hip extensor spasticity walked at an average speed of 41 m/min whereas persons with either moderate or severe spasticity averaged only 17 m/min ($p < .001$). Similar velocity differences were seen as a result of knee and ankle spasticity.

Significantly more hip flexion (26°) was achieved by persons with absent hip extensor spasticity than those encumbered with moderate (18° of hip flexion, $p < .05$) or strong (15° of hip flexion, $p < .05$) hip spasticity. The presence of severe ankle extensor spasticity also was associated with decreased hip flexion ($p < .001$). Swing knee motion was influenced by knee, ankle and hip extensor spasticity.

Selective Motor Control

Only two patients were able to activate Grade 3 selective strength in their hip flexors. These two persons walked at a velocity of 56 m/min in contrast to the 21 m/min averaged by patients with Grades 0-2 hip flexor strength ($p < .01$). Grade 3 or better knee flexor strength was obtained by three persons who were able to achieve significantly greater velocities (55 m/min) than those with weaker knee flexor grades (19 m/min, $p < .001$).

Different grades of selective hip flexor strength did not result in statistically different amounts of swing hip motion but did influence swing knee motion. Persons with Grade 3 hip flexors displayed more knee flexion in swing (55°) than those with Grades 0-2 selective control (28°, $p < .05$). Grade 3 or better knee flexor strength also led to more swing knee motion (49°) than did Grades 0-2 strength (29°), however this finding did not reach statistical significance.

A pertinent correlation between selective muscle strength and gait velocity was found only by combining all muscles studied (summing the muscle test scores of the flexors and extensors of hip, knee and ankle). The best relationship ($r = .72$) was defined by expressing muscle grades in Beasley quantitated values that were established by measuring torques produced by persons with varying selective grades. When selective muscle test scores were assigned their common arbitrary values (Grade 1 representing 20% of normal strength; Grade 2, 40%; Grade 3, 60%; etc.), the summed score displayed a correlation of only .58 with velocity.

Upright Motor Control

Two persons with strong UMC hip flexion grades averaged 45 m/min, six with moderate grades achieved 34 m/min, whereas 11 patients with weak grades walked at only 18 m/min. Having either a moderate or a strong hip flexion grade resulted in a significantly higher velocity (40 m/min) than that achieved with a weak grade (18 m/min, $p < .005$). Persons with either moderate or strong UMC knee flexion grades also walked at higher velocities (35 m/min) than those with weak grades (19 m/min, $p < .05$). The rate of hip flexion on the UMC test displayed a correlation of .75 with velocity achieved.

UMC grades were less strongly associated with differences in swing motion. Persons with strong or moderate hip flexion grades had 24° of swing hip flexion while those with weak grades averaged only 17° of hip flexion ($p < .05$). UMC scores at hip and knee were unrelated to swing knee motion.

Patterned Muscle Strength

Two persons were able to activate their hip flexors strongly in pattern and 10 displayed moderate strength. One of the individuals with strong hip flexor grades was a "poor walker" because of severe spasticity. In general, however, persons with moderate or strong patterned hip flexor strength achieved significantly greater velocities (35 m/min) than did individuals with poor pattern strength (15 m/min, $p < .05$). Moderate to strong hip flexor grades were also associated with improved swing hip and knee flexion; persons with stronger grades averaging 23° of hip flexion in contrast to the 14° of hip flexion demonstrated by the persons with weak grades ($p < .001$). Swing knee flexion values were 36° and 20° respectively ($p < .05$).

Three persons exhibited strong knee flexion in pattern. These persons achieved significantly greater velocities ($p < .05$) than those with weak pattern strength. Swing knee motion was not influenced by pattern strength of the knee.

DISCUSSION

In terms of predicting gait velocity to be achieved, selective Grade 3 flexion at the hip or knee identified the best walkers, all having velocities of 50 m/min or greater. (A companion study to this project has indicated that velocities at this level are associated with unlimited community level ambulation in the SCI population.) Selective strength grades less than 3 were not predictive of gait velocity. All but one of the four individuals with Grade 3 hip or knee flexors also demonstrated an absence of spasticity. The "good walker" with spasticity was able to utilize the less involved limb for substitutive maneuvers during gait.

The challenge then is to differentiate among the less able persons, those who would be most likely to respond to FES. The UMC and patterned muscle tests were able to distinguish intermediate from poor walkers; persons with moderate or strong grades on these tests achieved significantly greater velocities than those with weak grades. Thus these tests identified a significant muscle reserve not detected by selective muscle testing that is available for FES supplementation. In the presence of strong spasticity, velocities in individuals with moderate/strong UMC or patterned muscle test grades were reduced to less than 20 m/min.

The finding that spasticity exerted a strong influence on velocity and motion achieved confirmed a previously established selection criterion of excluding highly spastic individuals from FES gait assist (3). Kralj, however, has found that FES may be of benefit in reducing tone in some spastic individuals (3). Further research is needed to document the effects of FES on spastic muscles during gait and determine whether spasticity is a contraindication to FES gait assist.

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8.4

A CLINICAL APPLICATION OF ELECTROTHERAPY FOR ACCELERATING WOUND HEALING

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ABSTRACT

Accelerated wound healing has been an observed effect of low intensity direct current (LIDC) in the range of 200 to 800 microamperes, but presently available electrotherapeutic equipment does not provide this particular range of stimulation. A small and portable LIDC stimulator was specially constructed and utilized in a study of the LIDC effects on wound healing rates of thirty inpatients with indolent ulcers at the Spaulding Rehabilitation Hospital. The patients in each treatment group were paired by age, diagnosis, wound size and wound etiology. Comparison revealed 1.5 to 2.5 times faster healing in those receiving LIDC. In addition, there was significantly less debridement, no wound infections and more resilient healed scars. LIDC appears to be a convenient, reproducible and efficacious method for improved healing of chronic open wounds and warrants more widespread application for the clinical setting.

INTRODUCTION

There has been an increasing interest in alternative methods for promoting wound healing, especially for indolent wounds. These particular wounds tend to be a frustrating aspect of patient care management for the health care team, physician, third party payers and most all patients. There is presently a myriad of products that offer some possible alternative but as yet most have been less than satisfactory for these types of wounds.

The use of electrotherapy has been suggested to provide beneficial and reproducible healing even when all other methods have failed. Assimacopoulos first reported using a low intensity negative electric current of 75 to 100 uA at the ulcer sites.[1] He concluded that this current accelerated healing and caused a stronger scar tissue to form. Wheeler, Wolcott and Hardwicke described the benefits of using low intensity direct current (LIDC) for a wide variety of wounds.[10] A standardized procedure for applying the LIDC was formulated by these authors. The protocol called for a current of 200 to 400 uA for normally innervated skin and 400 to 800 uA for denervated or decentralized skin to be applied at the wound site. The negative polarity was to be at the wound for a 3 day period or until the wound was found to be aseptic. Following this period, the positive polarity was applied at the wound site. The negative polarity was then utilized as the ground electrode located 25 centimeters proximal to the wound site. There was a total of 6 hours of stimulation given in 2 hours segments. These periods of stimulation were separated by a 4 hour "off" time that accounted for a total of 14 hours for the LIDC application.

The researchers reported healing rates of 2 to 3.5 times faster than their controls along with stronger tensile strength of the scar tissue and anti-bacterial effects in contaminated or infected wounds. The protocol was repeated by several researchers and similar acceleration of healing rates, stonger scar tissues, and bacterial effects with no harmful effects to the patients were described.[2,4,5,6,7,8]

The use of LIDC electrotherapy warranted consideration in an area of clinical use which desperately needs predictable benefits for healing indolent wounds. However, the LIDC protocol and the bulky equipment needed to be redesigned in order to maximize its practical application in the clinical setting. There were many aspects that had to be considered before establishing the LIDC protocol in a clinically operative form, such as patients schedules, staffing and time constraints. The LIDC stimulator had to be specially constructed to accurately provide the current and still be small, portable and easy to operate. It was hoped that these revisions would maintain the similar beneficial aspects noted in earlier studies. A study was organized to collect wound healing rate data and investigate the effectiveness of this revised LIDC protocol for the practical clinical setting.

MATERIALS AND METHODS

Patient Population

A total of 30 inpatients were considered for participation in this investigation. The subjects were paired according to age, diagnosis, wound etiology, location and approximate size. One member of the pair was randomly assigned to receive the LIDC protocol while the other would continue with conventional wound therapy. This would consist of wet to dry gauze dressing, different solution soaked dressings, whirlpools (hydrotherapy) or a variety of other techniques. Both wounds were debrided prior to admission to the study for recording healing rates and consent forms signed.

Equipment

A small, portable electrical stimulator was specially constructed utilizing the latest in integrated circuitry (Fig.1).[a] The unit accurately stimulated at 100 uA increments with only a small variance of ± 10% and utilized a 9 volt battery as a power source. Audible and visual alarms were built in for continuity of the prescribed current level. It also contained an internal 2 hour timer which would cease the output current and sound an audible alarm.[b] The unit possessed a rugged exterior and a plastic clip for attaching to the patient's clothing.

The electrode material was modeled initially after the stainless steel mesh electrodes in earlier studies. The various electrodes were later modified from flexible carbon material much more resistant to corrosion and breakage. The material was gas sterilized at 40°C prior to exposure to the wound dressing. Each electrode was destroyed after the individual patients completed the study.

Study Design

The scheduled treatment application of LIDC required 2 hours of stimulation repeated twice daily. This was performed 5 days a week. The 2 periods of LIDC stimulation were separated by a 2 to 4 hour pause during which the unit was in the "off" position. The unit was connected in the morning and would automatically turn off within 2 hours after initially turning the unit on. It was restarted again in the afternoon following a 2 to 4 hour pause during which the unit was in the "off" position.

The wounds, debrided prior to admission to the study, were irrigated with saline solution and packed daily with either saline damped gauze or various absorption gels. The electrode was then placed over this conductive interface so that it was not in actual contact with the wound surface. A waterproof tape or adhesive transparent dressing was applied over the dressing to maintain moisture and keep the electrode in place (Fig.2).

The electrode at the wound site was termed as the active electrode and the electrode 15 to 25 centimeters proximal was termed the ground electrode. The active electrode was the negative polarity for the first 3 days of the LIDC application. The positive polarity acted as ground located in its respective position. Following this 3 day period, the polarities were reversed such that the positive polarity was now at the active electrode and the negative was acting as the ground. It is important to note that when the negative polarity was utilized as the ground, the electrode size had to be at least twice the size of the active electrode to avoid possible irritation to the skin. The positive polarity was then maintained as the active electrode until the wound healed or a plateau in healing was noted. If such a plateau was reached, the protocol of negative polarity at the wound site for a 3 day period was restarted.

Current output levels were set between 300 and 500 uA for patients with normally innervated tissues and 500 to 700 uA for denervated or decentralized skin. The current level could also be empirically assessed by using this general rule: the current level was too high if a bloody exudate appeared, and too low if copious serous drainage resulted.

Wound healing progression was recorded for both groups on a weekly basis. Area measurements of length, width and depth to within the nearest millimeter was performed by the nursing staff and recorded without prior knowledge of earlier measurements. Weekly photographs were also taken with a Polaroid Land Autofocus 660 camera. Data collection continued for 5 weeks or until the patient was noted as healed.

RESULTS

There were a total of 30 inpatients in the investigation and Table 1 provides a general description of the study population with regard to the paired groups. The patients that used the LIDC application showed a 1.5 to 2.5 times faster healing rate when compared to their paired controls. The overall healing rate was twice as fast for the LIDC group in addition to the other beneficial effects of this treatment. Table 2 shows the overall healing mean on progression for both groups. One example of accelerated healing noted in the LIDC group is shown in a sequence of three pictures taken in two week intervals after the LIDC protocol was initiated (Fig. 3).

An additional effect noted in the LIDC group was a stronger scar tissue noted upon palpation of the wound closure and the original wound margins. Highly contaminated wounds became less colonized and no wound infections occured with either the stainless steel mesh or the flexible carbon electrode materials. The LIDC group did not require any further debridement once the wound was properly debrided at the onset by either surgical or enzymatic methods. There was an overall subjective response of decrease in the pain and discomfort that usually accompanies an open wound when compared to the control group. This was unsolicited by the investigator or the health care staff. In addition, there were no harmful effects noted in patients during the application of LIDC.

The control group presented a slower healing rate with respect to the paired LIDC group. The scar tissue showed the lines of contraction typically seen in a healed wound. The healed tissue appeared thin and fragile, and at times would have reopened at follow up outpatient re-examinations. The control wounds would typically redevelop eschars that required repeated debridement as often as every two weeks. This resulted in a considerable amount of pain and discomfort in addition to what the patient was already experiencing while the wound was slow in healing. Many of the control patients were discharged home with the eschars still over the unhealed wound site.

DISCUSSION

The LIDC protocol and equipment adapted for the clinical setting demonstrated the similar beneficial effects to those reported in earlier literature, especially the acceleration of healing rates. The day-to-day use of the equipment and protocol was easily carried out by the nurses or physical therapists and sometimes even the patients. The dressing technique and materials were compliant and required the same amount of time to apply as did the conventional dressings. The specially designed LIDC unit was both small, portable and simple to operate by the staff and patients. The unit was conveniently attached to the patients' clothing and was connected to the wound-ground electrodes by thin insulated wires usually found with TENS units. Thus it provided an ease of movement by the patient while maintaining the accurate current dosage of LIDC. The staff and patients did not find the LIDC unit to be a hindrance during activities of daily living or their participation in therapy programs.

The beneficial effects of a stronger scar tissue and bacterial influences were important in the clinical setting. Sometimes these factors become significant when considering discharge or possible complications of patients during their already extended stay in the hospital. The subjective response of decreased pain and discomfort experienced by the LIDC group was unexpected and unreported in prior research. This positive effect could possibly be explained by the suspected analgesic effects of direct current stimulation.

This mode of electrotherapy should be considered for the clinical settings especially when considering the reproducible benefits of the LIDC application, non-harmful effects, and ease of clinical treatment and operation. Judging from our experience with LIDC, it could become the treatment of choice for treating indolent ulcers rather than the fruitless experimenting with methods that show little improvement of healing. Otherwise, it would remain frustrating to ourselves, third party payers and most of all, the patient and families who must endure pain and discomfort and possible amputations. The technology and materials are presently available to seriously reconsider this beneficial use of electrotherapy.

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WEIGHT TRANSFER TRAINING USING BIOFEEDBACK AND ELECTRICAL STIMULATION

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ABSTRACT

A training system has been designed for persons suffering from impaired gait function due to cerebrovascular accident or trauma, to recover static and dynamic standing balance and weight transfer skills that are fundamental to gait performance. The device uses electrical stimulation and video graphics with audio cues as biofeedback during the training exercises. Patient responses are analyzed locally by the device with results available on CRT and/or printed and plotted with an integral thermal graphics printer and can also be stored on tape for later transfer to a host computer for offline analysis.

INTRODUCTION

Many persons who sustain a cerebrovascular accident or an injury to the head are initially unable to walk or are able to walk only with a great deal of assistance. These persons may also be left with impaired function of their hand and arm. For these individuals, extensive gait and upper extremity training is required to assist them in achieving their maximum functional potential. During the months following the initial trauma, the patient will spontaneously recover many of the motor skills. The therapist will try to prevent further debilitating effects such as disuse atrophy, abnormal movement patterns and joint contractures while attempting to accelerate the recovery process during the treatment period.

Preliminary results (1) which examined gait and balance parameters in adult hemiplegic subjects, before and after a four week intensive rehabilitation program, indicated that subjects improved by learning to rely more on their less affected limb. This strategy leads to the development of abnormal movement patterns and an asymmetrical gait. Specific parameters which did not improve with rehabilitation were: percent of body weight taken thru the cane, and percent of body weight borne thru the involved limb during quiet standing as well as lateral and anterior-posterior weight shifting onto the affected limb. It was therefore proposed that a specialized weight transfer training program using performance feedback and electrical stimulation be implemented during rehabilitation in order to facilitate the incorporation of the affected limb into balance and gait activities.

BACKGROUND

Biofeedback has been used both to encourage and to limit weight bearing in stroke patients. One such device is the Limb Load Monitor (LLM) (2). The LLM consists of a pressure-sensitive transducer built into a shoe insole and a control box worn on the belt which emits an auditory signal whose pitch is proportional to the pressure applied to the transducer. This type of device has utility in gait training, but its usefulness is limited by the accuracy and reliability that can be achieved in any kind of transducer that attaches to the foot of a person with abnormal gait.

A training program was evaluated at Rancho Los Amigos Hospital which combined biofeedback of joint position and electrical stimulation of paretic musculature to facilitate wrist extension in hemiplegic patients (3). The equipment included an adjustable table to support the arm, a visual and auditory display unit, an electrogoniometer and an electrical stimulator. A patient was seated at the table with his arm stabilized to isolate wrist motion. Following a signal from the display unit to initiate exercise, the patient performed a maximal voluntary extension effort. The display unit provided continuous auditory and visual feedback, proportional to the amount of wrist motion. After the patient achieved a pre-set threshold angle, a success light was displayed, a repetition counter was incremented and electrical stimulation was applied to obtain a full joint extension. Following an appropriate rest period, the patient was signalled to make another effort.

A group of 30 hemiplegic patients was randomly divided into two groups, one which received positional feedback stimulation training and one which did not. With the wrist positioned in 30 degrees of flexion, average isometric extension torque in the control group increased 0.13 Newton meters while the study group improved an average of more than 1 Newton meter. This finding represents a 70% increase over the study group's initial torque of 1.5 Newton meters. Both control and study groups started the program with nearly equal average torques.

The concept of positional feedback and electrical stimulation has also been applied to the lower extremity with success (4). A study of 20 control and 20 study patients demonstrated a 67% increase in knee extension torque for study patients compared to a 260% increase for the control patients. Active range was also significantly greater for those patients treated with positional feedback and electrical stimulation at the knee.

SYSTEM DESCRIPTION

The Standing Feedback Stimulation Trainer consists of a platform containing two forceplates that measure vertical forces only, a control consol, custom microcomputer, and an audio/visual display system for feedback of information about the distribution of weight between the two forceplates, and a structure to provide support for the patient and forceplates.

The audio/visual display is mounted on a stand directly in front of the patient. A regime of feedback similar to the successful method of treating motion of the wrist, elbow, and knee is used (3).

The video display consists of a monochrome 512 X 416 pixel video graphics system upon which various animated stick figures and graphics symbols will impart position and balance information to the patient. For example, lateral shifting of weight from side to side is depicted by a stick figure standing with legs spread apart on a 'teeter-totter', the angle of which is proportional to the relative weight distribution of the patient.

The forceplates measure 12.5 inches wide by 25 inches in length and are supported at the corners by aluminum beams. Attached to these beams are metal foil strain gauges, each of which is capable of detecting vertical forces in the range of 0-50 pounds with a linearity error and hysteresis of 2%. The strain gauge amplifiers for each plate are located just in front of the plates to reduce electrical noise. The eight outputs of the strain gauge amplifiers are sampled by the microcomputer via eight sample and hold amplifiers whose outputs are sequentially measured by a 12 bit analog to digital converter (ADC). This configuration eliminates drift in potentials measured due to the conversion time per channel required by the ADC and essentially captures a slice in time for the forceplate activity. The data collected may then be used to determine total force and also quite easily the location of the center of that force for determining lateral and fore-and-aft motion about the plates.

The microcomputer houses the analog sample and hold circuitry, ADC, video graphics controller which can generate vector and character commands, a tape controller for a small digital cassette tape drive, power supplies, and a thermal graphics printer which is able to print and plot the results of the treatment runs. Command entry for the system is made via a video terminal operated by the therapist.

The tape drive is used to record the actual raw data collected by the system for later analysis by a larger computer system. A serial port is provided for transfering the data stored on tape to the larger computer. Initial projections are that data from the force plates will be sampled at 50 Hertz and that the data will be stored in memory for trials lasting up to one minute. At the end of a trial, a data record is written to the tape and then another trial can begin. This recording scheme can be altered to store only resultant fore-and-aft and lateral body weitht shift percentages to reduce the data storage requirements and thus to extend the trial time or sample rate. Concurrent sampling and storage on tape is a possibility as well.

The therapist is able to take a weight reading using the control unit and enter the percentage of body weight to be used as the goal thresholds for loading of the involved limb. Based upon the goal threshold values, with and without stimulation, automatic sensitivity scales are set for the feedback display unit. The control unit can also be used to read out previous efforts of performance by the patient and monitor how much of the weight relieved by the non-involved leg is being taken by the hand for stabilzation (differences between the sum of the two forceplates and total body weight).

A dual channel stimulator is used which the therapist can set stimulus amplitude levels to be used for stimulation of the quadriceps, gluteus maximus, medius muscles, or gastrocnemius used for facilitating voluntary effort. The stimulator is turned on and off by the microcomputer according to rules established by the therapist at setup time. Several options are available, such as on and off at specific balance percentages, etc.

TRAINING PROCEDURE

The goal of this project is to provide the therapist with a clinical tool to improve the patient's standing balance, weight transfer skills, ability to maintain weight bearing on the involved limb, and ability to control ankle plantarflexion. Specifically, it will focus on patients with the following problems:

1. Those who have ambulation potential, but who cannot begin gait training because they cannot bear weight on the involved leg and/or do not have adequate balance while standing.

2. Those who are ambulating, but a) have a grossly asymmetrical gait due in part to their inability to transfer and maintain weight on the involved limb and/or b) are hesitant about placing the foot of the involved leg ahead of the opposite foot because they are insecure about transferring their weight onto the involved limb in that position.

3. Those who are ambulating but have inadequate plantarflexion control in mid- and/or terminal stance.

4. Those who are ambulating, but have an unstable gait due to inadequate balance.

The trainer provides the use of the following six tasks for training static and dynamic balance:

Task 1: Patients lacking ability to laterally transfer weight to the involved limb will stand with one foot on each forceplate. They will be instructed to shift weight from the uninvolved limb to the involved limb each time a 'start' light appears on the feedback display unit. As they do so, information about weighting the involved limb will be continuously displayed. For those patients who are unable to shift all of their weight onto the involved limb and hold that position for a short time, electrical stimulation of the hip and knee extensors will be initiated near their weight bearing limit. Stimulation will provide facilitation and support for the patient to continue loading the involved limb. The system will record the number of repetitions of the task and the maximal degree of weight transfer with and without stimulation.

Task 2: The patient will be asked to stand with one foot on each forceplate and their weight evenly distributed between the two. The task is to maintain an equal distribution for a time period established by the therapist. Visual and auditory feedback will be used to indicate to the patient that he is leaning to the right or left. Electrical stimulation of the appropriate hip extensor and abductor muscles will be available as an option to provide an additional cue to the patient that he needs to activate specific muscles to resume a posture of equal weight distribution. Various parameters, such as the number of times the patient exceeded various levels of assymetery or the integral of the absolute value of the deviation from equal weight distribution, will be automatically calculated and available in printed form.

Task 3: With one foot on each forceplate, the patient will be asked to shift his weight from one limb to the other to track a moving signal on the screen of the display unit. A command signal will move at random or according to a predetermined pattern from side to side, similar to tasks performed by previous training devices (5,6). A second signal will be controlled by the patient. Shifting his/her weight to the left foot will cause the signal to move to the left and vice versa. Their task is to shift their weight to follow the command signal as closely as possible. Electrical stimulation will be used as in Task 2, as a cue, when the patient is too far out of range of the command signal. Parameters such as the integral of the absolute value of the difference between the two signals will be recorded.

Task 4: Patients who have difficulty in maintaining stability at the knee and hip of the involved limb during gait while weight is transferred from the uninvolved limb to the involved limb will be positioned so the foot of the involved limb is on the forceplate while the foot of the uninvolved limb is positioned posteriorly. Their task will be similar to that described under Task 1, except that they will now be shifting their weight forward and backward instead of laterally.

Task 5: To improve plantarflexion control during mid-stance, the patient will be fitted with a knee brace on the involved side fixing the knee in 20 degrees of flexion. Both feet will be positioned evenly with the involved side metatarsals on the forceplate but the heel on the floor to the rear of the forceplate. A padded positioning bar will be placed against the calf musculature to fix the tibia position and limit the ankle from going into more than 10 degrees of plantarflexion. Patients will be instructed to plantarflex their foot each time the 'start' light appears on the feedback display unit. Information described under Task 1 will be indicated on the feedback display unit. When the goal threshold is reached, electrical stimulation of the gastrocnemius will be applied, thus facilitating their voluntary effort.

Task 6: To improve balance during single limb stance, the patient will be asked to stand with one foot on one of the forceplates while sliding his other foot forward, backward and to the side. Stimulation of the hip and knee musculature of the involved extremity to provide adequate support will be used as necessary. Feedback will be used to show the patient how much weight is being supported by the support limb during the task. When the patient becomes sufficiently skilled, they will be asked to maintain their balance on one limb while the opposite foot is lifted off the floor.

SUMMARY

The weight transfer trainer system described above utilizes successful techniques of Biofeedback coupled with electrical stimulation in a series of structured training tasks to recover static and dynamic standing balance and weight transfer skills in persons suffering from impaired gait function due to cerebrovascular accident or trauma. On line analysis of patient performance as well as the capability of detailed offline analysis suggest that this type of device could play an important role in the rehabilitation process.

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9.1

Limitations of Augmentative Communication Systems in Progressive Neurological Diseases

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ABSTRACT

This paper presents a case study of an individual who at 44 began noticing the effects of amyotrophic lateral sclerosis (ALS). Recommendations addressing the increasing communicative needs of this individual were developed and implemented; however, training and use of the systems suffered. This paper analyzes the observed breakdowns of augmentative communication system use and the effects that technology and personal attitudes might have on the eventual functionality of augmentative communication systems.

INTRODUCTION

Recent technology has provided severely disabled individuals with a means of communicating their needs. Even individuals with very limited motor control are often able to access communication systems with sophisticated interfaces, allowing them to generate messages with either printed or voice output. Problems exist when the individual's disability involves a progressive degeneration of the motor system. Over time, systems originally recommended to suit the individual's needs become less functional or non-accessible because of the severity of their disease.

BACKGROUND

The case discussed here was originally referred to the Assistive Device Center for assistance with his communication. At that time this patient was diagnosed as having amyotrophic lateral sclerosis (ALS). Due to respiratory complications, he had a tracheostoma and could not speak. He communicated by writing messages and occasionally pointing to pictures.

Given the severity and the progressive nature of ALS, the recommendations developed allowed for progressive loss of motor ability. Two options which addressed the need for an augmentative communication system were proposed. First, a microcomputer-based communication system which would allow him to select entire words or spell his messages was recommended. Word processing capabilities would also be included. Initially, the client could access the keyboard directly and later, through a system of flexible interfacing, he could access a single switch used for keyboard functions when standard keyboard use was no longer possible. Option two provided various communication systems including a hand-held typewriting device such as a Canon Communicator (Telesensory Systems Inc.), or the Sharp Memowriter EL-7100, or the Zygo Communicator-matrix communication device (Zygo Industries). Switches to be activated by the hand and/or head movement were recommended in the event that the Zygo Communicator was implemented. These two options addressed the need for functional expressive language; one that avoided "yes-no" answers to questions posed by others, thus limiting his choices and messages. In addition, the recommendations assumed that the client would lose the functions necessary to operate the initial interfaces recommended (a keyboard accessed with his hand); requiring the implementation of alternative selection modes (a single switch activated first with gross hand movement, then with head movement).

Three years later the client was referred for a reassessment of his communication needs. During those three years a Canon Communicator was purchased; however, at the time of the referral, he had been unable to use the Canon for several months due to lack of hand control. His communication consisted primarily of "yes-no" responses; no other means of spontaneous message initiation were observed, and he could not voluntarily signal an attendant or family member in emergency cases. Due to the progression of ALS, his only reliable movement was raising his eye-brows. A scanning communicator (Zygo Scanwriter) which allowed selection and printing of letters by sequential activation of the brow wrinkle switch (Prentke Romich, Inc.) was recommended.

The relative sophistication of this second recommended system provided the client with the necessary technology; enabling him to access a system with very limited residual motor control. The communication system recommended here challenges current commercially available technology, in its most sophisticated form. In addition, the recommendations address the nessessity to match specific client needs to commercially available devices (1). The solution attempted to solve such complications as the variability between the client's physical abilities, the complexity and rate of change of technology incorporated into assistive devices and the wide range of devices for various disabilities.

After a three month period for acquisition of the recommended system components, device recommendations were implemented and training for system use was initiated. After five one-hour sessions, the client failed to demonstrate a functional use of the system. He did not attempt to initiate, nor did he attempt to respond to his audience. Explanations addressed to analyze this dilemma cover a broad range of topics.

DISCUSSION

First, the client was without a communication system for approximately nine months from the time that use of the Canon was no longer possible and a new system was available for his use. The time without a functional means to communicate probably contributed to a dependent role termed "learned helplessness". Even given extensive training, the client remained largely dependent upon others to assist his communicative needs, taking on a "respondent role", never initiating communication (2) . Learned helplessness appears to be the result of the client experiencing things over which he has no control (i.e., in this case, the extended time period without a communication system). Later, even when he could control the situation, he remained passive and did not initiate actions to control his environment.

In an attempt to determine other causes for his non-communication, we used a series of yes-no questions. The client indicated that he "had nothing to say". Topics began to shrink, requests to plan the day's menu for lunch and dinner were not even complied with. These types of topics had been of significant interest to him when using his Canon communicator.

Analysis of the interpersonal relations between family members was also considered. Previous communicative interactions with members of his family are not known; thus, it could be possible that very little interaction took place. Additionally, certain personalities assume a very reserved role. He might have not been the "talkative" type.

The demands of the technology used may also have played a role in this situation. The client was fitted with the most sophisticated communication system available. Scanning, as a selection mode is a complex process and is cognitively more demanding than alternative modes (i.e., direct selection) (1). The physical selection mode consisted of the brow-wrinkle switch which was the only available interface given the client's demanding needs and residual motor control. As time wet on, even this switch became unuseable, and an infrared ocular switch was procured (Words+). The difficulties in setting up and adjusting this switch further contributed to problems of use and may have resulted in even less use of the overall system. All the features contained in this system may have had a negative effect on the client's approach to system use, as well.

CONCLUSIONS

Based on our experience with this client and others using similar technology, we can offer these general conclusions:

1. Provision of an augmentative communication system, in cases of adults who previously had speech, must be timely. We have observed that clients who were provided with systems as soon as possible were able to avoid the "learned helplessness". These systems have often been very simple word boards employing manual scanning. These clients also become better users of electronic systems.
2. Consideration must be given to communication needs of the client and the family. Interpersonal relationships may foster dependence and each family member's agenda may differ. Since severely disabled clients are dependent on able-bodied family members or care providers to "set them up" on the system (e.g. attach the switch), and to initiate communication with them, this may be a deterent to successful use.
3. We must view the entire augmentative communication process from the client's point of view, rather than a stereotypical model of successful communication. The client may not wish to communicate or to interact. If so, we should respect that position.
4. Even in cases such as the one presented here, clients have a right to make their own decisions regarding system use. Without the procurement of a system, this client would not have been able to make a decision as to the cost/ benefit tradeoffs in using it to communicate.

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COMMUNICATION MAPPING FOR SEMANTIC COMPACTION SYSTEMS

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ABSTRACT

A form of artificial intelligence called semantic compaction can enable communication aid users to output sentences and word groups rapidly. To exploit this technique requires that an individual's communication needs and wants be viewed in a systematic manner termed semantic mapping. This paper supplies the rationale for such mapping as well as giving a basic guide to the clinician or aid user concerning how to proceed. This will apply to high functioning individuals as well as those who are mentally retarded.

Semantic compaction, whose commercial name is MINSPEAK, is a type of concept keyboard, which, though having a restrained number of images (about 60 is average) has the capacity to define whole sentences in 5 or fewer keystrokes. The system is now in use in over 100 sites across the United States, Europe, and Australia.

Semantic stands for "meaning". Compaction stands for "shortening." Semantic compaction is a new technique in relation to computers; it operates on the level of human meaning. It associates an image illustrating a concept--such as food, maleness or femaleness, time, or heat -- with a key.

The therapist or user programs the device by typing in full sentences using letters. He or she then selects a sequence of concept keys -- usually 3 or 4 --to represent that sentence. From then on, the user can access the entire sentence with only 3 or 4 keystrokes.

The opportunity that semantic compaction offers the users of a communication aid is potentially revolutionary. If an aid user were to have at his disposal hundreds of sentences designed for the many contingencies of daily life, he or she would be able to have something to say on many occasions and would not have to go through the laborious task of composing utterances through letters or word abbreviations. But, how to compose such a list of sentences?

If severely handicapped individuals are going to get the most out of a communication aid or use it at all, in some cases, they must be able to output utterances quickly and easily enough for them to appreciate the power of language output. With semantic compaction, they can do this. Whole utterances couched in appropriate language can be output by striking several or even a single key. But, how does one compose such a list of appropriate utterances that will peak the interest of users?

While many clinicians are quite skilled at giving their clients and patients the elements of communication --that is, some spelling skills and/or some word assembly abilities --it is quite a different task to look at the overall communication patterns of a person's life.

The reasons for this are various and complex. First, as communicators themselves, clinicians suffer from certain natural blind spots. People are often their own worst observers. Second, the science of linguistics has not developed fully its branch of semantics. This has had effects all down the line in scientific research. One of these effects is that there is no means to predict absolutely a person's semantic needs.

What this paper will propose are a series of considerations concerning semantic mapping. Perhaps a definition of semantic mapping is appropriate here. By it, we mean a schedule or paradigm that will allow a clinician to look at a person's communication needs and desires in a more systematic and/or predictive way.

We shall proceed in an anecdotal manner at first in order to capture some of the realities of communication aid use. The following demonstrates what happens with a traditional aid that does not use speakers but inexperienced at the specialized needs a device user encounters. Namely, needs for language can be: quickly retrievable, appropriate, productive, flexible (fits many situations), individualized (clever, insightful, proper, bright, witty, zestful), potentially easily modified.

Semantic mapping offers exactly what the term suggests. A direction through the seemingly endless "jungle" of vocabulary selection, phrase and sentence sequencing to arrive at a systematic means of expressive development with a communication device.

Choices are much easier if delineated. Targeting communication environments and proposing a possible 10 to 50 phrases and sentences within the environment provides a much needed organization to the task of choosing.

Now the clinician can review the environments using her personal knowledge of the client and make the initial determinations.

1. What environments does the client presently interact within.
 A. What possible phrases or sentences would produce the greatest "pay-off" for the client within these environments immediately.
 B. What possible future additions within these environments might be realistic goals.
2. With careful planning for implementation, what new environments would have positive developmental potential.
 A. Selection of the number and types of new environments would be based on careful analysis of the client's total program.
 B. Linguistic and conceptual development would, by necessity, be greatly interwoven within these goals.
3. What practice activities could be integrated into therapy sessions and classroom tasks.
 A. Determining a realistic number of new phrases for a "3 week period" to be practiced in a drill fashion.
 B. Setting forth skits and role playing scenes for interactive practice.
 C. Planning field trips to "test" the new-found skill.

Ironically, once the clinician has set forth a plan of this type, the individualizing can then easily be accomplished. Presenting the "Semantic Map" to the client with further delineated choices allows the client involvement on an appropriate level.

For example, let's say the clinician has chosen the communication environments: Greetings, Partings, and Introductions as the most immediate "pay-off" areas for the client. Using the Semantic Map as a guide she explains to the client the rationale and expectations within these choices. She further explores social implications within the different items under each. Now, the client can make choices, with clinician counseling, such as, "that phrase is longer - you risk losing listener attention; that one is proper anywhere - but rather ordinary. You could take them both - but that increases the number of new items to remember and you would have to use the right one in the right situation." (One cannot help but be reminded of verbal language development and pragmatics that go on within mother-child modeling and development).

The procedures we have been discussing are now coming into focus because semantic compaction is allowing for the rapid and appropriate output of so many different sentences.

This process is radically different from abbreviation systems where a sequence of letter or numbers is taken to represent sentences. For instance, "idu" can be programmed to stand for "I don't understand". However, after 50 or more sentences, it becomes hard to make the letter codes bear any relation to the words in the sentences. As the codes become more and more arbitrary, users begin having trouble recalling what sentences they have stored and under what codes they have stored them. Semantic compaction, on the other hand, allows the user to operate on the meaning level at all times.

If they have participated in a semantic mapping of their life situations, they can have a large linguistic repertory to express across a wide variety of situations in a timely and individualistic manner. Clinical support materials for semantic mapping are available from the Schneier Communication Unit in Syracuse, NY; Crippled Children's Hospital School in Sioux Falls, South Dakota; the Prentke Romich Company of Wooster, OH.

semantic compaction or semantic mapping. Envision a scene in which an augmentative communication device user, Donna, is visiting a physician's office The purpose of the visit is for a routine physical examination and to request a support letter for vocational training. The doctor has not seen Donna for two years and is unfamiliar with her use of an augmentative communication device.

Donna needs to impress the doctor that she has sufficient functioning to warrant investment in vocational preparation. The bulk of this responsibility lies with her communication.

The attendant assists Donna into the doctor's office and they wait. As the doctor enters he is asking the attendant, "How's Donna doing, it's good to see her again - just exactly what's the problem with her today?"

The attendant smiles and motions to the communication device, saying, "Well, she's pretty good doctor, we just came to get this routine physical exam form filled out. Donna wants to tell you something here - herself - here...."

The doctor stops and looks at the device asking, "Well, well, what is that thing -- some sort of computer, huh? What's she use that for - school work and such?"

"Yes and to communicate," the attendant answers.

Since the doctor's arrival Donna has been working diligently with her communication device. She is now watching the attendant and the doctor and as the doctor begins to rummage through her chart saying, "Well, let's see, we probably better look in your ears here." She presses a button and the device says, "I'M O.K. I'D LIKE TO TALK TO YOU ABOUT MY FUTURE." Doctor: "Oh, it talks too - whaddit say?"

Donna has the machine repeat, and the doctor looks to the attendant, who says, "Oh, she's just telling you we came here to get you to write her a letter of support for vocational training."

Doctor: "Oh, yeah - like what do you have in mind, I mean is she thinking there's something she can do - like where would she be going? You know, I've known her since she was a tad..."

Donna still is diligently typing into her device, it finally says, "PLEASE LET ME TELL YOU I WANT TO GO TO MARSHALL COLLEGE."

The scenario finishes with the doctor confessing he is very short on time - maybe he should just do the exam and the attendant can call his nurse and let her know what this is "all about."

Let's suppose we could replay that scene. As the doctor comes into the room, Donna makes 2 strokes on a device using semantic compaction, and it says,

"HI HOW ARE YOU?" 2 more strokes, "I AM USING THIS DEVICE TO TALK TO YOU. PLEASE LISTEN CAREFULLY. IT WILL BECOME EASIER TO UNDERSTAND WITH PRACTICE."

The doctor stops and says, "My God, whaddid that thing say?"

Donna has the machine say it again. The aid points to the liquid crystal display where it is printed out. 2 strokes, "I NEED YOUR HELP. 2 strokes, I FEEL FINE. 2 strokes, I NEED YOU TO WRITE A LETTER FOR ME. 3 strokes, THE LETTER WILL BE TO STATE VOCATIONAL TRAINING."

The doctor says, "Yeah, o.k., what am I supposed to tell them?"

3 strokes, "I AM HEALTHY. 3 strokes, I WANT TO TO GO MARSHALL COLLEGE, 4 strokes, TELL THEM YOU THINK IT IS A GOOD IDEA."

The difference in these scenes is not the people involved. It is the application of semantic mapping and semantic compaction.

A familiar scene in therapy sessions is a clinician requesting a client to "tell" her what he wants programmed into the memory portion of the communication device. There inevitably ensues a "down period" in which each does nothing. Then the clinician frequently will begin to propose possible random items; i.e., "Well, how about your name, your dorm number, your favorite TV shows, your friends' names."

As unflattering as this may be to the professional, when analyzed from an objective view, it is not unexpected. Clinicians are experienced, interactive

9.3

A MODEL FOR INTERNATIONAL EVALUATION OF AUGMENTATIVE COMMUNICATION DEVICES

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A model to promote high quality, efficacious communication devices by providing organized feedback to users, clinicians and those who develop and manufacture devices is presented. An internationally organized system for assessing devices is proposed which includes evaluation of such factors as: safety, reliability, ease of modification, input requirements, storage capabilities, output specifications and clinical efficacy ratings. Devices would be evaluated in clinical settings with the information being collected regionally, then nationally and internationally, with dissemination occurring internationally through existing networks.

BACKGROUND

The field of augmentative communication has shown tremendous growth in the past five years. The introduction of technology to this field has had a major impact both in terms of the development of dedicated communication devices and the implementation of commercially available mass-market devices (including hardware and software) - both modified and unmodified to meet the needs of disabled users.

Consumers, professionals, developers, manufacturers, and funding agents are confronted with a critical lack of standards and information in this area. A model for international evaluation and information exchange regarding augmentative communication devices is proposed.

Although various protocols for device evaluation have been developed, (most recently by IPCAS*) these protocols have not gained wide acceptance and have not been implemented with sufficient structure to have an impact on the field.

The proposed model presents a framework for international cooperation in information exchange concerning communication devices. It is hoped that such a model will assure that non-speaking individuals can avail themselves of the most appropriate and current technologies to meet their needs.

FACTORS

To implement an international model for device evaluation, one must consider a concise protocol which is easily completed, is as objective as possible, and which lends itself to easy interpretation and transmission.

Certain key factors and a measurement scale are presented. Parameters to be evaluated will include:

- safety
- reliability
- durability
- ease of modification
- input requirements
- storage capabilities
- output specifications
- clinical efficacy ratings.

A numeric scale to accompany these parameters will assure easy interpretation, and utility across different linguistic and cultural contexts.

IMPLEMENTATION

Regional

The depth of evaluation for any device will be determined by whether or not it is a device new in concept, or an updated version of an already evaluated device. All devices will be field tested, with evaluations made by the user and by multidisciplinary clinic teams, including both clinicians, engineers and technicians.

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*International Project on Communication Aids for the Speech Impaired.

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Ratings will be carried out by at least two independent clinics. The concept of having clinics specialize in an area of device evaluation such as voice synthesizers, or computerized systems will be explored. There will be a coordinating panel responsible for informing clinics of new devices and in assisting clinics in obtaining devices for evaluation. The panel will also accept and review the reports, make decisions about further testing when clearly conflicting results are reported, provide feedback to the manufacturer or developer, and ensure that the evaluation material is passed to similar panels in all other participating jurisdictions.

International

If standardized tools for evaluating the equipment and reporting that evaluation are adopted, then equipment and accessories whose clinical efficacy is not highly dependent on cultural context could be assessed in any region with the results enjoying international validity. Time and dollar savings are potentially large using such a scheme.

Information could routinely be made available first through telenets like CONFER which now links Sweden, England, the United States and Canada. Because very significant development is taking place so rapidly, first line communication must keep pace, if the advantages of that development are to reach the user in a timely fashion. Detailed information could be disseminated through existing resource publications such as Communication Outlook, Communicating Together, Augmentative and Alternative Communication or The Non-Vocal Resource Book (Trace Centre).

Conferences and meetings would provide regional and international opportunity to present and discuss the ongoing evaluation of such a system for evaluating communication devices. The authors would be prepared to coordinate such an evaluation.

It is proposed that one or several international organizations such as ISO, IPCAS or ISSAC be approached to oversee the development of this system and after a three year trial, if the project is a success, assume ongoing responsibility for its coordination.

Further, the notion of formal acceptance or endorsement of a device will be discussed, as will the implications of unfavourable reviews. Stress will be placed on the importance of the cooperative nature of the model - one that promotes high quality, efficacious devices by providing organized feedback to developers and manufacturers, and to users and clinicians.

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9.4

THE EFFECT OF INCREASED SENSOR AREA ON INPUT RATE AND ACCURACY OF THE EXPRESS III COMMUNICATION AID

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ABSTRACT

By enlarging the effective sensing area of an EXPRESS III handpiece, the input rate of some users increased by as much as 1.6 times of the rate using the standard handpiece. The large sensor allowed for placement of the handpiece anywhere within a square to select the square. It also had a hystoresis effect which helped to increase the accuracy and thus the rate for spastic individuals.

BACKGROUND

The EXPRESS III system, made by the Prentke Romich Company, has several input modes. One of these is a direct selection technique where an individual selects squares by pointing to the LED in the corner of the square with a hand-held sensor. The EXPRESS III uses an orthoganal array of LEDs (light emitting diodes) which sequentially flash on and off. Each LED is in the center of a 1" square, and emits light through a 3/8" diameter clear window. The sensor (a photo diode) senses the light from an LED in the same manner as a lightpen works on a TV screen to determine the square being selected by the user. If the LED remains sensed and lit for a set period of time (which is user adjustable), that square is selected. In order to make another selection, the sensor must first lose the signal from the original LED, and then sense another LED, even if the same square is going to be reselected.

PROBLEM BEING ADDRESSED

Clinical evidence shows that the sensor on the EXPRESS III can be too small for effective use by some individuals. The small size of the sensor reduces the speed at which a person may select a square, since the handpiece must be placed very accurately. For a person who has very slow fine motor control, or who is spastic, this can be very frustrating. If the user places the sensor very close to the LED target, but not on top of it, the selection will not register until the sensor is moved the very small distance to bring it directly over the LED. For some individuals, moving the sensor a small distance, and keeping it there, is very difficult, if not impossible. What usually happens is that they overshoot and select another square. One solution to this problem would be to enlarge the LED target. It is impractical, however, to increase the LED size for all of the LEDs. The most logical solution was to enlarge the effective area of the sensor.

MATERIALS AND METHODS

The effective area of the sensor was increased by taking the sensor and mounting it in a black tube, with an inside diameter of 0.75", about 0.75" above the bottom of the tube. The tube had a protective clear plate set at the bottom. The sensor was mounted to another clear plate at the top using a microminiature phono jack. This allowed the center of the sensor to be anywhere within a 1 1/8" diameter area around the LED. This arrangement is shown in a corss-sectional view of the enlarged sensor (Figure 1).

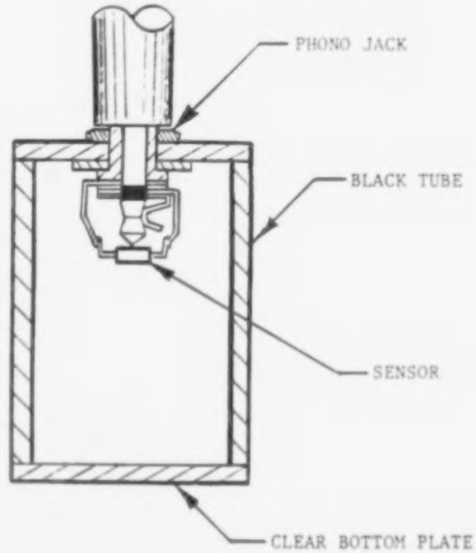

Figure 1

PRELIMINARY TESTING

Three subjects and one control were used to test this new design. All were in wheelchairs. Subject 1 had mild cerebral palsy; Subject 2 had moderate to severe athetoid-ataxic cerebral palsy; and Subject 3 had moderate to severe spastic cerebral palsy. Subject 4 was a non-handicapped control subject. The subjects were asked to select 8 squares in a specified order, and to do this twice in a row, so that a total of 16 squares were selected for each test run. After several preliminary runs, each subject was asked to repeat the test run three times. The raw results, of the time to complete a run and the number of wrong selections, were recorded. The preliminary runs were used to allow the subjects to become familiar with the required selection sequence, and to be near the top of their learning curve. The results of the test are shown in Table 1.

Subjects 1 and 3 increased their output rate by 63% and 37% respectively. Subject 2 showed a decrease in speed of 13%. The control subject showed an increase of only 7%.

DISCUSSION

Ease of Use

For all of the subjects, the physical ease in using the large sensor and the small sensor were about the same. The subjects expressed no difference between the two, and no visual evidence was present to show that using one sensor produced more physical stress than the other. However, the prototype was physically larger and made of an opaque material. This larger sensor obstructed the user's view of the squares more than the small sensor. All of the subjects complained that it was more difficult to tell exactly where they were pointing when using the large sensor. Even with this problem, however, Subjects 1, 3, and 4 said they felt it was mentally easier to use the larger sensor.

Speed

The increase in speed is a result of the larger sensing area. The subjects did not have to be as accurate with their pointing in order to select a square. There was greater improvement for the subjects who had more difficulty with small corrective movements. For the normal subject who had accurate pointing and good small motion control, the advantage of the large sensor was not as great for speed, but was still advantageous for ease of use.

Accuracy

There appears to be no difference in accuracy between the two sensors. However, for Subject 2, there was much difficulty in knowing where the center of the sensor was pointing when using the larger sensor. This was accentuated due to the fact that this subject used the EXPRESS III in a high position, where the angle of sight was quite low.

Height Selection

The sensor became active during the downward motion at 1.2", and inactive on the upward motion at 1.7", nominally. The height of the sensor above the surface was chosen to be 0.75" since it was desired that the sensor be active when placed on the surface but inactive when over an inch above the surface. The height and inside diameter combination made the largest effective diameter of the sensor roughly 2.3" at its maximum distance from the surface. This was larger than desirable. A better design would have a constant effective diameter at all heights. After making the modification, it was discovered that the increased admittance of light made the sensor active at 2.5" but inactive at 3.5". However, this did not seem to pose significant problems with the design since the subjects kept the sensor flush with the surface during use, and lifted it infrequently.

Subject #	Small (spm)	Errors %	Large (spm)	Errors %	Increase (X)
1	27.7	0.0	45.1	1.9	1.63
2	25.4	2.8	22.0	10.8	0.87
3	30.9	3.1	42.3	0.0	1.37
4	64.0	0.0	68.6	0.0	1.07

spm = selections per minute
% = # wrong selections / total # of selections
X = spm of large sensor / spm of small sensor

Table 1

Diameter Selection

The diameter of 0.75" was selected based on the distance between the effective area of the LEDs. The minimum distance between the effective area of two LEDs is 0.625". To produce a hysteresis effect when moving back and forth between neighboring LEDs, the effective diameter of the sensor needed to be greater than this minimum distance. By choosing the diameter of the tube as 0.75", the center of the sensor has to be more than 0.56" away from the center of the square before being able to select a neighboring square. To reselect the same square again, the sensor has to be less than 0.44" away. There is, therefore, a hysteresis of 0.12" on the radius.

CONCLUSION

It is apparent that for some users, a larger effective sensor area would greatly increase the ease and speed of input without any loss (and in some cases an increase) in accuracy. The modification to the original design is minor, and would not significantly increase the cost. Another version of the sensor will be made to try to solve the problem of visibility and non-uniform effective sensing diameter. The results of this second version will be reported later. Once an acceptable design is reached, it will be passed on to the Prentke Romich Company.

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DEVELOPMENT OF AN AUDITORY SCANNING COMMUNICATION SYSTEM WITH MULTIPLE VOICE OUTPUT FOR SEVERELY DISABLED USERS

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ABSTRACT

A prototype communication system using the DECtalk voice synthesizer for auditory message scanning has been developed for visually impaired, motor disabled, nonspeaking persons with reduced cognitive skills. The concept of auditory scanning has been optimized based on the DECtalk ability to produce multiple, distinct, and highly intelligible voices for speech output. Individual voices are paired with different system functions so that multiply disabled users can learn to operate a device with varied auditory feedback. A unique DECtalk option that allows a user to custom design his own voice retains the personal and egocentric aspect of communication in this system that often is overlooked in device design.

INTRODUCTION

Speech synthesizers that have been incorporated into communication systems for nonspeaking persons generally assume the role of speech prostheses. That is, the synthesizer produces voice output for expressive communication (1). A less common use of the speech synthesizer is to present instructions and a language representation system to the user for device operation. For the visually impaired, nonspeaking person, the application of speech synthesizers for the latter purpose is obvious. The person who cannot process visually presented material on a CRT or communication board can operate a device solely through the auditory channel.

A microcomputer-based communication augmentation system for two nonspeaking, physically handicapped persons with severe visual impairment has been reported by Beukelman et al. (2). In the two systems described by the authors, visually impaired individuals received single letter audio feedback as they formed messages with Morse Code or an alphabet scanner. For the individual with added cognitive deficits, a less demanding language representation system is needed. The introduction of an auditory scanner which presents message menus to the user may be appropriate for the visually impaired nonspeaker with reduced cognitive skills.

DEVICE DESCRIPTION

The auditory scanner was designed using the DEC Rainbow microcomputer, color monitor, printer, DECtalk voice synthesizer, DEC graphics utilities, and ADC-1 data acquisition and control system. The ADC-1 system includes a BSR X-10 environmental controller as well as an interface for a single switch.

There are three major components to the auditory scanning program. Since the DECtalk has the unique capability of producing multiple, distinct, and intelligible voices, a different voice is used for each component. Initially, the auditory scanner introduces itself to the user and describes how it works. The device presently functions as a traditional scanner with only one menu. Thus, the Device Introduction component instructs the user to listen to a number of messages and close a switch when the desired message is heard. Since a cognitively impaired patient may forget where the switch is, the instructions may include a statement indicating the switch location and procedure for switch closure.

The second component, which is spoken in an entirely different DECtalk voice, is Message Presentation. Pre-designed messages are presented to the user along with simple, color graphics. At this stage, the user simply familiarizes himself by listening to the entire message menu. The cognitively impaired individual with memory deficits is given repeated opportunities to learn the menu prior to selection everytime the device is rebooted.

The final component is Message Selection. Here, the messages and simple graphics are repeated at a slower rate with a longer interstimulus interval. The messages are presented with the voice that is used in the second component. There are two possible outcomes of this component. The user may fail to close the switch. At this point, the system (in voice #1) indicates to the user that he has not selected a message. The device also says that if the user wishes to hear the messages again, he should hit the switch.

The second outcome in the Message Selection phase is that the user closes the switch appropriately to select a message. The device, then, speaks out the message in a third DECtalk voice. This final voice, which is produced louder for the communication partner, is custom-designed by the user. The DECtalk voice synthesizer is flexible enough to create idiosyncratic voices which can represent the nonspeaker's personal expression. This function should increase the pragmatic aspect of augmentative communication. For the first time since the inception of synthetic expression, the user actually has control of his speech production. Instead of speaking in a Votrax or Echo-like voice, the individual can modulate his pitch, rate of speaking, and tone. It is hypothesized that this option will increase device use and improve conversations between nonspeakers and their communication partners. After the selected message is spoken, the device (in voice #1) tells the user that the menu will be repeated if the switch is closed again.

If the switch is not hit again, the device remains silent for a fixed period of time. At regular intervals, the user is informed that the device is waiting, and can be reactivated by closing the switch. This function assists the cognitively impaired patient with memory deficits who may, in fact, forget that the system is running. The reminder will be heard at these intervals as long as the device is running.

CLIENT DESCRIPTION

The prototype auditory scanner was designed originally for one 25 year old male who presented with 2 year S/P anoxic encephalopathy. The patient is nonvocal with no consistent means of expression. On neurologic examination, he exhibits severe spasticity bilaterally, left hemiparesis, and frontal lobe dysfunction. EEG shows severe diffuse cortical dysfunction. Hyperactive reflexes are evident.

Visual skills are severely compromised. Ophthalmologic examination reveals adequate visual acuity. Perception and discrimination can not be assessed due to the lack of a consistent motor response. Visual evoked responses are normal. The presence of a visual agnosia was confirmed on neuropsychological testing. He demonstrates severe visual field cuts, as well. Vision appears to be his most effected sensory modality.

Auditory skills appear intact. The patient can identify musical tones and localize objects auditorilly by turning his head in the direction of a sound. He appears to demonstrate adequate comprehension of language though formal assessment can not be accomplished because of output limitations. His ability to learn and retain new information is significantly reduced.

The patient can activate a single switch device with a headstick. Due to his reduced visual and motor skills and nonvocal condition, an auditory scanner was recommended (4).

At present, the auditory scanner contains one menu consisting of ten messages. The messages were chosen by the client, his family and clinical team. Half of the messages are for communicative interchanges. For example, the device will produce, "Please scratch my back."; "What time is it?"; "How are you today?"; and "I'd like a drink of water." The remaining five messages are used for appliance control. Since the system includes a BSR X-10 environmental controller, it will turn on/off a fan, radio, and tape recorder. The program is written so that a selected message, such as "It's hot in here. Please turn on my fan.", is changed during the menu repetition into, "It's cold in here. Please turn off my fan." Once the patient demonstrates that the system is appropriate for his cognitive and communication needs, additional messages will be added.

DEVICE APPLICATION

As the number of patients presenting with head trauma and severe disability continues to increase, rehabilitation engineers and clinical teams are being asked to design augmentative communication systems for multiply involved patients. The number of nonvocal patients presenting with visual impairments and/or cognitive deficits is rising, as well. Therefore, multiple-function augmentative communication systems, such as the device described above, are becoming necessary.

The auditory message scanner with DECtalk voice synthesizer is unique in a number of ways. First, the auditory scanner offers the visually impaired patient a means to receive information for device usage solely through the auditory channel. The use of synthesized speech for device control increases the clinical utility of speech technology for the multiply disabled user.

The DECtalk ability to produce multiple, distinct, and intelligible voices should increase device usefulness, as well. Individuals with reduced cognitive skills can learn to use the device by associating different voices with the various system functions. The high quality of the synthetic speech also will assist the patient who presents with compromised auditory processing skills.

Institutionalized, multiply impaired patients often spend extended amounts of time alone. The auditory scanner gives these individuals opportunities for continual man-machine interaction and dialogue without the need for communication partners. The availability of a device that "speaks to" the patient, rather than simply "speaking for" the patient offers additional language stimulation for rehabilitation. The added feature of environmental control can help the visually and cognitively impaired individual affect his daily surroundings through automated auditory control.

Since the user of this new system will custom design his own voice for message expression, his motivation for device usage will increase. The pragmatic aspect of augmentative communication should improve. The pragmatics of augmentative communication is a component of system design that often is overlooked by device designers. Individuals who investigate the pragmatics of nonvocal communication often ask why nonspeakers are not using their devices optimally. They have suggested that slow communication rate and partner dissatisfaction with devices negatively affect communication interactions (3). The factor of poor quality, impersonal speech output often is neglected in the pragmatic analyses. The introduction of the custom designed voice for alternative expression should, indeed, positively affect the use of devices and the value of communicative interchanges for system users.

The system described here is of limited use due to its restricted number of messages. Auditory message scanners that use directories and syntax nodes for multiple menus can be designed for the visually impaired, nonspeaking patient who can process large amounts of information. The simple auditory scanner can be made more complex as a cognitively impaired individual regains processing abilities. In addition, the order of messages on the traditional scanner can be changed according to frequency of usage. If the device can count how often a message is selected, then it can adjust the order of presentation of the menu items to optimize on communication rate.

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9.6

MODULAR TECHNOLOGY FOR THE SEVERELY PHYSICALLY DISABLED
THE 'LAINEY' SYSTEM

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ABSTRACT

This paper presents the evaluation results of Federal Contract # 30-83-0283 awarded to the Prentke Romich Company with a subcontract to the University of Tennessee Rehabilitation Engineering Program. The system developed is a modular component system which provides appropriate mobility, seating, control and interfacing with a communication aid and computer. With only one input mode required the system allows the severely disabled to compete on a more equal footing in a broad range of life situations.

INTRODUCTION

There are thousands of school age children in the United States who are being excluded as active participants in their education process. These are individuals whose intellectual abilities are within or above normal levels, but who are so severely physically handicapped that they cannot voluntarily use their body parts to speak, write, manipulate objects, sit, or move independently.

Technology does exist today to meet the needs of moderately involved physically handicapped students. Joysticks and single switch modes are just two of a variety of switch options for operating powered wheelchairs and some computers. There are keyguards which assist the student with direct selection of computer keys or expanded keyboards for those with only gross motor placement skills.

For the more severely involved, the problem resides in their inability to access available technical aids, even with single switches. These students often have only their head control to call on for function. This one body part must be able to operate a powered mobility device, a communication aid, and also the input to a computer system. Any control system developed must be portable and have multiple functions if it is going to maximize the very limited motor skills.

The rapid development of technology does not in any way guarantee compatability of technical components which an individual may require. It would be quite surprising to find compatability between computer, mobility and communication aids in terms of interconnecting plugs, chargers, and electrical operation. Often, available input interfacing on the equipment does not make it possible for the individual with a severe disability to independently change modes of operation.

BACKGROUND

Contract # 30-83-0283 was awarded the Prentke Romich Company to join together presently existing technical components into a compatible modular system. The Lainey System, as it came to be known, allows the severely physically disabled to have access to powered mobility, computer operation, and speech via an augmentative communication aid.

The result of the project is the Lainey System (1). The prototype was fit in January, 1985. The evaluation component began with the receipt of funds and has been ongoing.

EVALUATION PLAN

There are three facets of the evaluation plan:

1. Client Profile
2. Pre/Post Performance Data
3. Technical Evaluation

The Client Profile provides an overview of physical, functional, psychosocial and intellectual abilities of the user. The goal is not to develop a comprehensive set of client data but to provide a profile of an appropriate user for this particular project. Fitting to only one client was allowed so the client profile describes a seventeen year old lady with tension athetosis (Lainey). She has no functional use of any body part with the exception of her head. In order to use her limited head control she must be seated in a very aggressively supported seating system and her arms must be restrained below a tray (2). She attends school in a special education environment in an academic program. She lives at home with her mother and teenage sister. Both home and school are accessible. The school bus and home owned van accommodate a powered wheelchair. Long range goals include college and independent living.

Lainey has worked with the UTREP for over nine years and has exceptional patience for prototypes. She also puts considerable physical and emotional effort into making technology work for her.

Performance data reflects the functional skills acquired which are a results of fitting with the Lainey System. Areas assessed include:

1. The ability to maintain an upright posture with and without an appropriate positioning system.

2. The ability to independently operate a powered wheelchair. Data reflects distance traveled, number of sites visited, cost effectiveness of powered mobility (independent mobility versus attendant costs).

3. Physical operation and functional outcome of an augmentative communication system were monitored. Data collected reflects speed of access, length and quality of language content, environments and relationships in which communication has been facilitated. Language samples were taken with strangers, knowledgable professionals, and family members. Data collected reflected comparisons using no augmentative aid (only gestures, twenty questions, etc.), using an Etran and using the Minspeak.

4. The ability to access computer hardware and software in educational work stations at school and home have been evaluated.

Finally a report has documented the technical aspects of the Lainey System. Included are suggestions for production models based on the Lainey fitting. Compatability of components and marketing plan to include compatable modular components are stressed.

DISCUSSION

Several general issues of interest were made evident during fitting and evaluation of the Lainey System.

1. Through the cooperative effort of two facilities (UTREP and Prentke Romich Company) over a thousand miles apart, existing technologies can be married into a truly modular system.

2. Considerable technical and therapeutic resources must be available during the provision of advanced technology. Fittings should be augmented by clear instruction manuals but they, in themselves, are not enough.

3. Not even seasoned professionals can anticipate consumer reaction. Fitting of at least one prototype unit to a consumer in a real life setting is invaluable both in terms of product and process evaluation.

CONCLUSION

The fitting of the Lainey system to one client has been completed. A full report of evaluation findings and the technical report will be available after March 30, 1985 at which time the contract ends. Also available at that time will be a 15 minute video tape documenting the functional use of the Lainey System. Both written materials and video tape will be available through the Prentke Romich Company, 1022 Heyl Road, Wooster, Ohio 44691.

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9.7

A PORTABLE DECTALK SPEECH SYNTHESIZER

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ABSTRACT

A portable version of the high quality DECtalk speech synthesizer has been developed by the Trace Center at the Waisman Center, University of Wisconsin-Madison. The portable DECtalk can use two internal 12 VDC batteries, an external 19 VAC adapter, or an external 24 volt DC power supply. It can be turned on and off either manually or automatically by the user's communication aid via the RS 232 serial communications port.

INTRODUCTION

Speech synthesizers can play a very important role in the education and employment of non-vocal severely handicapped individuals. However, high quality speech synthesizers with unlimited vocabulary have not been available in a portable form. To be of complete use for handicapped people, a speech synthesizer must be easily portable and operate for at least a full day on a single charge.

BACKGROUND

The DECtalk (made by Digital Equipment Corporation) is a commercially available text-to-speech synthesizer which uses advanced speech synthesis algorithms to produce very intelligible speech. It has seven different voices: three men's voices, two women's voices, a child's voice, and a user definable voice. This is an important capability, since many children and women refuse to have a male voice speak for them regardless of how sophisticated and intelligible the synthesized speech sounds.

The DECtalk can be directly connected to a modular phone jack, so the speech can be transmitted over telephone lines without losing its quality. The DECtalk also has a dictionary that can have new words added or have a whole dictionary downloaded from a host computer. It has a very sophisticated algorithm for dealing with numbers, and also has the ability to produce the alternate pronunciation for certain words.

These features make the DECtalk an excellent speech synthesizer for non-vocal individuals, and it was therefore chosen as the speech synthesizer to be made portable.

DESCRIPTION

Design Goals

The overall design goal was to make the DECtalk portable and usable by someone in a manual or electric wheelchair. This meant making it mountable to a wheelchair, resistant to damage from rough use, and resistant to damage from being used in the rain or around areas where accidental liquid spillage might occur. The DECtalk must also be made able to operate for a long time on a single charge, so that it will not be necessary to frequently recharge the unit. The minimum requirement was that the charge should last a full day. For ease in adjusting the volume and adding an external speaker, an external volume control knob and a speaker jack were added to the outside of the package.

Operational Modes

The portable DECtalk operates in two modes: a manual mode and an automatic mode. In the automatic mode, the custom designed power supply automatically turns power on for the DECtalk mother board as soon as the DECtalk receives a character via the RS-232S serial port. It is ready to receive text after another 250 ms, and turns off after a specified time-out. The automatic mode saves battery power by turning the DECtalk on only when text is being spoken and automatically powers down the system after it is finished with the utterance. This increases the length of time the DECtalk can be used before needing to be recharged.

The manual mode operates in exactly the same way as an unmodified DECtalk. It is turned on and off by a switch (i.e., the DECtalk is always on when turned on by the switch). This mode is necessary to allow the portable DECtalk to work with software expecting an unmodified DECtalk.

Automatic Power Down

The power board has two adjustable time-outs, one to turn off the fan and the other to power down the DECtalk mother board. The time-out for the mother board can be varied using jumper pins from 1/2 second to 256 seconds, and the time-out for the fan can be varied from 16 seconds to 256 seconds. In order to prevent possible overheating of the mother board, the fan will not turn off until after the

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mother board is off.

Internal Battery Operation

When turned on manually, the DECtalk can operate for about 5 hours using the internal batteries. In the automatic mode, the DECtalk can operate for 1 to 5 days without being recharged, depending on how much is spoken. It takes ten hours for the unit to be fully recharged. While being recharged, the DECtalk can still operate using the external 19 VAC adaptor, or an external 24 VDC power source.

Electronic Hardware

The heart of the power board design is the switchable 25 watt DC/DC converter (RIFA Model PKA 2231). It supplies the necessary +5, +12, and -12 volts for the operation of the DECtalk mother board. A full wave bridge network allows the use of a 24 DC or a 19 VAC source to power the converter. The data line of the RS232 port, and the FIFO ready line of the speech output circuitry on the mother board, are used to enable the power-up circuitry. An RC timing circuit is used to reset the mother board after power up and a digital counter is used to perform the power down time out functions.

Packaging

The finished portable DECtalk unit weighs 22 lbs. It is 15.75 inches long, 12 inches wide, and 3.5 inches high. The casing is made of aluminum. The portable DECtalk is designed to be mounted onto the back of any standard Everest & Jennings manual or electric wheelchair using the standard E & J electric wheelchair controller brackets. The circuit board is fan cooled, and is shielded against emitting radio frequencies. The internal electronics are protected against damage from rain when mounted on a wheelchair. The switches to select the operational mode and the internal or external power source are mounted on the control panel. The control panel also contains a volume control knob, an external speaker jack, fuses for the power board and the internal batteries, connectors for the charger and external power source, and an RS 232 connector.

DISCUSSION

Initialization

Since the DECtalk powers up each time before speaking a phrase in the automatic mode, it will always be in the default voice of Perfect Paul, the standard male voice. If the user wants to use another voice definition, the input string must always have an initialization string which selects or defines the desired voice. This is also true with other variable parameters which are not retained after power down, such as speech rate.

Serial Communications

Special attention needs to be made regarding the RS 232 serial communication from the host computer when in the automatic mode. The host computer must be able to send a character via the RS 232 port even though the DECtalk's RS 232 port is off. Some computers, such as the HX-20, have difficulty doing this. Also, the first character sent must be chosen so that it will not affect the spoken text if the DECtalk is already on. A space or the NULL character can be used. The host must then wait a minimum of 250 ms before sending the initialization string and text that is to be spoken.

Weight

The power requirements of the mother board are quite high, so large batteries are required to guarantee long operation as well as long life of the batteries. The batteries alone weigh 10 pounds. To support this weight, and to protect the circuitry from damage that could result from dropping the unit, the packaging was made from aluminum and weighs 6 pounds. The rest of the unit accounted for another 6 pounds, so the total weight of the system is 22 pounds.

Size

The limitations on how small the unit could be made were determined by the size of the mother board and of the batteries. The mother board is 10.5" X 15," and the batteries are 2.5" X 3.75" X 5.9" each, so the final size of 3.5" X 12" X 15.75" is about as small as possible (and is actually smaller than the orginal package from DEC).

Cost

The cost for a custom modification to a single standard DECtalk is approximately $2700. This modification includes $250 for the custom designed power supply, $2200 for packaging, and $250 battery, charger, and AC adapter.

CONCLUSIONS

The first unit has been placed with a non-vocal young woman who is using it daily at school on her E&J manual wheelchair. She uses a small external speaker in order to have the voice projected forward from the chair rather than sounding like it comes from the actual unit behind the wheelchair. She uses an Epson HX-20, a small portable lap computer, to input sentences to the DECtalk. She uses the predefined child's voice. She has been

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very pleased with the quality of speech and its portability. So far, the DECtalk has lasted an average of 7 days on a single charge when in the automatic mode.

AKNOWLEDGMENTS

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By Digital Equipment Corporation
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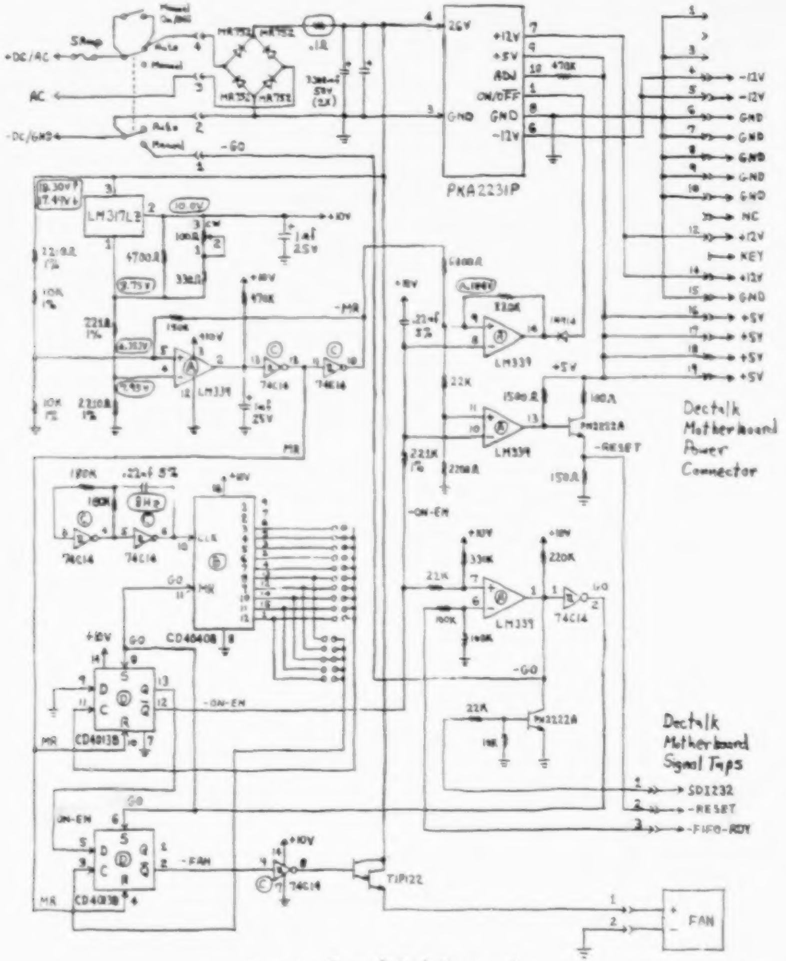
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-GO
MR752
.1Ω
330mf 50V (2X)
26V
GND
+12V
+5V
ADJ
ON/OFF
-12V
470K
PKA2231P
LM317LZ
10.0V
+10V
1mf 25V
4700Ω
330Ω
100Ω
CW
LM339
74C14
-MR
MR
180K
.22mf 5%
8Hz
CD4040B
CD4013B
GO
-ON-EN
ON-EN
-FAN
TIP122
PN2222A
1N914
-RESET
22K
221K
2200Ω
1500Ω
150Ω
100Ω
33K
220K
100K
-12V
GND
NC
+12V
KEY
+5V
Dectalk Motherboard Power Connector
Dectalk Motherboard Signal Taps
SDI232
-RESET
-FIFO-RDY
FAN

Dectalk Power & On/Off Board

DELIVERING ASSISTIVE DEVICES IN A REHABILITATION HOSPITAL SETTING : PATIENTS AND PATIENCE

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ABSTRACT

The paper describes some of the challenges and problems that have been experienced in establishing an assistive device service. The areas of critical importance are identified.

INTRODUCTION

A number of assistive device centres are being developed in various locations throughout the world. Experience has shown that the transition between conception of the idea and its realization can be fraught with many challenges, some anticipated, many not foreseen. The focus of this paper is to discuss some of the challenges experienced in the implementation of an assistive device program in a rehabilitation hospital, the strategies used in problem solving, and those areas which must be addressed if success is to be achieved.

BACKGROUND

The Glenrose Rehabilitation Hospital is funded by the province of Alberta and provides a wide range of rehabilitation services to the residents of Northern Alberta. The hospital has an adult and a pediatric rehabilitation unit with 288 beds and provides a school hospital program and outpatient services to approximately 5000 adults and children each year. The Assistive Device Service (ADS) was established in 1983. The core disciplines currently involved are speech-language pathology, occupational therapy and physical therapy. Active consultation services are provided by clinical rehabilitation engineering, education, physiatry, orthotics, social work and psychology.

CHALLENGES IN DEVELOPING A SERVICE

In developing the ADS it became apparent that there were several broad areas which required attention. These can be categorized under the headings of administrative support, philosophical development, knowledge base, and expectations.

Administrative support

The first challenge was to obtain administrative approval of the program. The initial proposal was developed by the director of the Department of Speech Pathology in 1982, having been preceded by many discussions with other departments. At a departmental level it was possible to obtain philosophical agreement of the need for such a program, provided that funding for additional staffing could be made available. Because of the economic climate the hospital administration were unable to approve additional staff positions. It was next decided to bring in a clinical rehabilitation engineer as a consultant for three days to provide: an intensive inservice on assistive devices for the clinical staff, an analysis of the hospital's needs and potentials in the provision of an assistive device service, and recommendations for the implementation of such a service including staffing, equipment and long term development. This strategy was successful in that it

a) convinced administration that it was appropriate and feasible to make a commitment towards the development of an assistive device service
b) convinced department heads that it would be feasible to reallocate staff assignments without adversely affecting other departmental services
c) provided a framework for the development of the service.

The second challenge concerned obtaining a broad selection of equipment that could be used in evaluating and training clients in the use of assistive devices. A broad selection of augmentative communication systems was available, however equipment such as electronic devices for mobility, environmental control, computer access and a broad selection of switches were not on site. Because of a strong commitment to the successful implementation of this service, Speech Pathology allocated a significant proportion of the department's operating and capital equipment funds to the purchase of this additional equipment.

The third challenge was to obtain dedicated space for the service. Considerable stress was placed upon both the team members and the equipment by the necessity to unpack and repack equipment from location to location. This was due to the extreme shortage of space in the hospital. It became a matter of utmost priority to locate an adequate dedicated space. Despite administration's sympathetic understanding of the situation, it nevertheless took over a year before a space could be assigned.

One of the continuing challenges in operating the ADS is to add gradually those additional disciplines which will improve the calibre of the service provided. Not only does this require intensive "lobbying" with the departments involved, it also requires modification of traditional service delivery procedures. As an example we were fortunate enough to obtain the services of the industrial arts teacher from our School Hospital as the ADS education consultant. However, because of the traditional structure of the school program, he was initially only able to attend meetings and clinics at times when his class was not scheduled (which translated into noon time meetings and little else). This restricted the contributions he could make to client management. Gradually over several months it has been possible to make arrangements that a substitute teacher will take over the Industrial Arts classes whenever a clinic has been scheduled. The next challenge will be to expand the mandate of which clients he can see. At the moment the mandate of the Education Department is to provide education to children attending the Glenrose School Hospital. However there are many instances when it would be invaluable to involve the education consultant with preschoolers, with children seen on an outpatient basis who attend other schools, or with adults who have needs in terms of educational and vocational access. Over time this expansion in the scope of his involvement may be possible.

Philosophical development
The philosophical development of the numerous disciplines involved in the ADS has been a process which has taken time plus considerable care and attention. The success of this process depends upon two factors: assigning an effective team leader and establishing basic trust among all team members. The particular professional discipline of the team leader is irrelevant, but success is far more dependent upon communication skills, organizational ability and a perceptive awareness of unspoken needs and fears of individual team members.

Initially there was a conscious effort to involve all members in all aspects of the client's needs assessment and system development. This was partly to facilitate knowledge acquisition as well as to provide mutual support in novel situations. Over time this process revealed those areas where there was overlap between professional roles and those areas where a need existed but no one claimed "responsibility". The team required assistance in dealing with the approach/avoidance issues which surfaced. These issues could be encapsulated by "don't touch my area, but I'm overwhelmed and can't do it." It took a year before it became evident that it was necessary to schedule time for philosophical discussions about roles and responsibilities. It was helpful to structure these discussions with precirculated agenda items, followed by the assignment of responsibility for implementation. As self confidence and skill develop, the time must then come when professional roles become more defined so that individuals contribute their expertise in those aspects wherein they have the greatest knowledge. This can be a painful process as members learn to survive without constant mutual support and learn to trust and develop confidence in their own as well as their colleagues' competence.

Much of the philosophical development centered around technology. There is a tendency to become fascinated by the attributes of equipment and therefore to develop preconceived notions of the system solution without paying sufficient attention to the client's needs and skills. It is imperative to have someone who will constantly bring the team back to the true issues under discussion. In our experience this constant refocusing upon the problem was provided by the consultant rehabilitation engineer and it has been a key element in the development of an effective service.

Another key element has been the involvement of the family in determining the priority of needs to be addressed. At the end of a clinic, the team coordinator summarizes all the recommendations, assigns specific responsibility for tasks (including a time frame) and provides this written summary to all team members and the family. A meeting is scheduled every second week to provide update reports for the clinical chart on each client currently involved with the ADS. This facilitates coordinated treatment and followup when more than one team member is involved. Assigning one team member as a contact person for each client has also facilitated communication. The contact person serves as an advocate for the client, family, and community, assisting in coordinating treatment and followup.

An interesting philosophical development has been the change from a 'departmental' orientation to a transdisciplinary orientation among team members. This has manifest itself in many ways from something as simple as the dedication of "department" equipment for relocation in the ADS room, to the reclassification of a speech-language pathologist's position to enable the hospital to hire a clinical rehabilitation engineer on a full-time basis.

Knowledge base
Establishing a core knowledge base about the characteristics of devices and the skills to effectively analyze client's needs and match them to device characteristics was achieved in several ways. The one component of an effective ADS team that was not available on site was a clinical rehabilitation engineer. Funds were allocated to bring in, on a frequent consultant basis, a clinical rehabilitation engineer who had considerable experience in working at an assistive device centre. The addition of this consultant was an excellent mechanism to educate the team in all aspects of assistive devices and client management. Another means of expanding the team's knowledge has been to make funds available so that they can attend professional meetings concerned with assistive devices. In addition, workshops have been held to not only educate team members on assistive devices, but to increase community therapists' awareness of assistive devices and their application.

When selecting the individuals who will serve on the ADS team, it is imperative to identify staff who are firmly rooted in their own professional areas, who are competent professionals, and who have a strong interest in working within a transdisciplinary team. Working with staff of this calibre is an on-going educational process for all team members, a process which has the potential to be intimidating but which is intensely stimulating. Having several staff on the team who have worked at the Glenrose in excess of a decade has also proven valuable because of their awareness of community resources.

Taking this ever-expanding knowledge base and applying it to the benefit of clients has posed many challenges. The team mustbe alert constantly to ensure an appropriate balance between medical versus educational versus technological versus therapeutical approaches. Situations arise where transdisciplinary activities may hinder rather than accelerate progress through the proposed plan of action. Such problems must be addressed openly and all team members urged to refocus upon the needs of the client rather than subsidiary issues.

Not all staff have the same competence in activities such as effective documentation. Basic forms and standards should be established early in the development of an assistive device service. As much of the documentation as possible is accomplished during team meetings or for pre-established time frames. This encourages the development of more even and complete reporting skills, and facilitates conformity in the interpretation of forms and procedures.

Expectations
Some of the less anticipated challenges centered around the development of appropriate expectations Each team member has found it necessary to work through a number of internal inappropriate expectations. These have ranged from "I'm only a .... so I can't learn to master that" to attempting to strive for a knowledge level that will be acknowledged as superior to all other members on the team. Inappropriate expectations of equipment include "when we get ... ", a tendency to be intolerant of equipment malfunctions, and impatience characterized by not taking sufficient time to learn to operate a piece of equipment competently. Inappropriate expectations of clients came at many levels. It took experience to learn that clients need to receive extensive training in the use of a system no matter how intelligent and motivated they appear to be,that it is unwiseto attempt to solve all problem areas for a client simultaneously, and that client selection is highly important particularly during the early phases of the program. Chose clients who have needs which can be dealt with quickly and effectively. This builds the team's confidence as well as demonstrates that the service is viable. Hospital departments and agencies in the community had inappropriate expectations that the ADS would provide instant "cures" and equipment. It is a continuous process to educate colleagues that the evaluation and training process is extremely lengthy and that followup will extend to infinity in many instances. The ADS has been established as a 'consultant' service and therefore the ADS staff do not "take over" the treatment of clients but rather consult with the treating team/therapists so that they may provide the ongoing training and support.

SYNTHESIS OF EXPERIENCES

There are some critical factors which should be considered when developing an assistive device service.

1. Staff selected for the program should be firmly rooted and competent in their own professional areas, should believe in the efficacy of transdisciplinary intervention and be knowledgeable in assistive devices and their implementation.
2. There should be strong administrative support, both in terms of financial backing, as well as commitment to persevere during difficult times.
3. There must be a significant investment in a wide range of equipment on site.
4. It is necessary to develop a thorough knowledge of community resources including service delivery and funding sources.
5. The service should evolve slowly, requiring considerable patience and caring at all levels.
6. Carefully select clients from a learning perspective and from a success perspective.
7. Technical input should be a working part of the team.
8. Field testing of equipment is an integral part of the process in fitting devices to meet client's needs.
9. If there is any component of the team not available on site, develop mechanisms to fund that aspect on a regular consultant basis.
10. Use the assistance of a person who has experience in developing an assistive device program in preparing the proposal for your service. If possible, use this person on a continuing basis to develop the knowledge base and guide the progress of the service.
11. Select an effective team leader, not from any specific professional discipline, but rather on the basis of their skills as a team leader.

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10.2

AN INTEGRATED APPROACH TO PRESSURE SORE PREVENTION FOR THE SPINAL CORD INJURED

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ABSTRACT

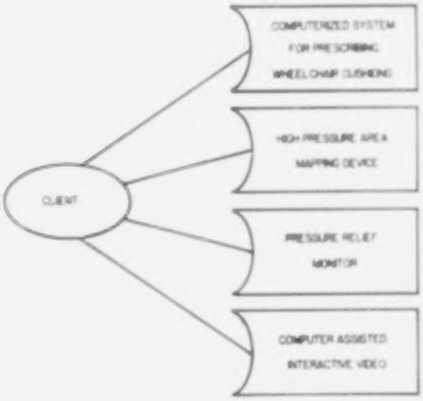
A multidisciplinary approach is used to prevent pressure sore formation. Through the establishment of a Pressure Sore Prevention Clinic, this is achieved by improving wheelchair cushion prescription procedures and by increasing the spinal cord injured person's awareness of factors relating to pressure sore formation.

INTRODUCTION

Pressure sores are a serious secondary problem for the spinal cord injured. It is estimated that every year, 14,000 spinal cord injured patients will be hospitalized for treatment of pressure sores (1). On average, an ischial pressure sore requires 78.8 days of in-patient treatment (2) at an estimated cost of $20,000 per sore (assuming $280 per diem cost). In addition to the financial burden, the person's personal, family, emotional, and social life is adversely affected. The pressure sore further restricts the person's lifestyle, often causing periods of unemployment, social inactivity, family problems and lack of self-esteem.

In 1972, Reswick and Rogers established the first pressure sore prevention clinic at Rancho Los Amigos Hospital. Since then, clinics have been established at Philpshill Hospital in Glasgow, Scotland (1976), at Royal Perth Rehabilitation Hospital in Western Australia (1976), at Conradie Spinal Cord Unit in Cape Town, South Africa, and at Texas Institute for Rehabilitation and Research (TIRR) (1976). These clinics have provided an effective way of reducing pressure sore incidence. For example, during the initial three years of operation, the Philipshill clinic reduced the number of patients admitted with pressure sores over the ischial tuberosities from 6.8% to 3.0% (3). The Perth clinic also reports a 33% decrease in admissions due to pressure sores (4), while TIRR has reduced the number of patients readmitted for pressure sore treatment from 31% to 4% (4).

Although each clinic follows the basic principles of individual prescription of wheelchair cushions, patient education, and scheduled follow-up appointments, the techniques, documentation, and instrumentation used, varies considerably between centers. In 1982, Helen Hayes Hospital began to establish a pressure sore prevention clinic in an attempt to bring together the strengths of existing centers to form a model pressure sore prevention program for spinal cord injured patients.


CLIENT
COMPUTERIZED SYSTEM FOR PRESCRIBING WHEELCHAIR CUSHIONS
HIGH PRESSURE AREA MAPPING DEVICE
PRESSURE RELIEF MONITOR
COMPUTER ASSISTED INTERACTIVE VIDEO

PRESSURE SORE PREVENTION CLINIC

Previous studies (5), had resulted in the development of a composite cushion concept, which was adopted for use in the seating clinic. A composite cushion may consist of several layers of different materials such as gels, T-foams, and high and low density foams, configured to give mechanical and physical properties in the composite that are superior to the individual material components alone. This design results in a versatile and effective cushion by minimizing the pressure under the buttocks and accommodating different body types, level of injuries, and personal preferences. In addition to the composite cushion, commercial cushions such as T-foams, Gels, Rohos, Jays, and specially fabricated Cutout Cushions are used. To aid in the fitting of cutout cushions, two ischial imaging methods are available in the clinic. The goal of the clinic at Helen Hayes Hospital, like the goals of the other centers, is the reduction of pressure sore incidence. This is achieved through an integrated program of cushion fitting, close follow-up evaluation, and education.

EXPERT SYSTEM FOR CUSHION FITTING

Particular effort has been directed towards providing a framework for transferring the expertise gained in our clinic to other rehabilitation centers. One approach gaining increasing credibility in private industry and medical diagnostics is the use of computer based expert systems. Expert systems are designed to:

a. make readily available the expertise of the specialist
b. provide a means for collecting and recording large databases for clinical research
c. automate part of the expert's task in order to increase productivity and lower costs (6).

In order to take advantage of these capabilities, an expert system has been developed at Helen Hayes Hospital to assist in the prescription of wheelchair cushions to provide guidance and knowledge for both the inexperienced and experienced therapist (7).

This program organizes the cushion selections and helps guide the evaluator in choosing an appropriate cushion with a minimum number of cushion trials. The patient's administrative data is entered, followed by a profile of his/her seating behavior, including skin conditions at the various sites, ischial prominence, frequency of push-ups and total sitting time. These characteristics then determine the acceptable or goal pressures that should be achieved at the various buttock sites. Finally, interface pressure values are measured on a three and four inch reference cushion. Based on these values, and clinical data the fitter is guided to evaluate either Foams and T-foams, composites, Cutouts, Gels, Rohos or Jays. The choice of subsequent cushion is based on the previous measurements taken. This process continues until the pressure goals, along with stability, temperature and comfort criteria are met.

Part of the program has an extensive scheduling routine which enables the therapist to schedule a follow-up appointment, usually six months later. These subsequent sessions are extremely important in preventing pressure sores. During the follow-up, cushions may need replacement due to ordinary wear and tear, or if the person's physical condition has changed, making his/her present cushion unsuitable, an alternative type or design can be provided. Also, the person's skin condition can be checked and documented. If an appointment is missed, the therapist is alerted the next time the program is used, thereby making it easier for him/her to follow-up patients.

An integral part of using the expert system is its potential to develop a large database. Not only can data be collected from the clinic at Helen Hayes Hospital, but the program can easily be implemented in clinics at other centers, thereby providing not only a transfer of expertise, but also a means to establish a multi-center database. Data collected could be used to determine which cushion types are most effective for different levels of injury, how long each cushion type can be used effectively before needing replacement, and the frequency with which cushions are prescribed at the various centers. The results gained would then be used to update and improve the clinical services by modifying the expert system and other clinical procedures.

MAPPING THE HIGH PRESSURE AREAS

One category of cushions frequently used in the seating clinic is Foam in which the high pressure regions under the ischial tuberosities are relieved by cutting out the cushion under these areas. The cutout is either left empty or filled with a softer foam. Success of this method depends on accurately locating the high pressure areas of the ischial tuberosities. TIRR developed a PEP-Pressure Evaluator Pad, a mat with a matrix of resistive paint which can provide this pressure information (8). At present, two less sophisticated methods are currently used in clinic. The first method is a modified wheelchair with an optical barograph, based on the pedobarograph concept (9). This arrangement causes high pressure areas to show up light and low pressure areas to show up dark. The areas are then marked on a removable grid and used to determine the cutout dimension. The second device takes an imprint of the person's buttocks. A frame with rollers covered by a canvas mat is placed in the person's wheelchair seat. Under the rollers is a special carbonized paper designed for gait analysis (Shutrak Walking Progress and Analysis Record) which provides a tracing of the buttocks' high pressure areas. The frame allows the therapist to lower and raise the patient onto the Shutrak without smearing the image of the ischial tuberosities. The patient first transfers on to the taut canvas, then is lowered until sitting firmly on the frame and Shutrak, and finally is raised and the tracing removed. This provides the therapist with a permanent tracing of the high pressure areas needed to determine the dimension of the cutout. In addition, this method is portable and allows the measurements to be taken in the users own chair.

PUSH-UP MONITOR

Research has shown that the risk of pressure sore formation decreases not only by reducing the pressure but also by periodically relieving the pressure under the buttocks, by leaning or pushing-up (10). In general, a pressure relief of five seconds duration every fifteen minutes is thought to significantly reduce the risk of pressure sores (11). To aid in the monitoring of push-up frequency of the spinal cord injured, a portable monitor has been developed to record push-up type, duration and time of occurrence. A similar monitor is being developed by Brian Andrews in Glasgow, Scotland. The monitor being developed at Helen Hayes Hospital is housed in an Ethifoam seat insert and connected to a force plate in the chair seat. The force plate is strain gaged, and with its accompanying electronics, measures the X and Y coordinates of the center of gravity, along with a Z component proportional to the total weight on the plate. These three signals can uniquely detect a full push-up, a lean to the left or right, and a lean forward. The monitor can operate for a month, thereby allowing for an accurate profile of the person's push-up behavior in his/her natural home environment. The results are then processed and displayed on an IBM PC. Therefore, patients who do not push-up frequently enough can be identified and be given additional instructions on performing push-ups and their importance. The effectiveness and retention level of the training can also be checked by subsequent monitoring sessions, with individualized modifications made where necessary.

COMPUTER ASSISTED INTERACTIVE VIDEO

In any clinic or rehabilitation program, emphasis is placed on patient education. An educational method is needed that actively teaches the principles of pressure sores, provides a measure of the comprehension level of the user, can be individualized to the patient's educational and comprehension level, and can be learned without an educator present all the time. Computer assisted interactive video, which combines the color and motion of video with the versatility of a computer, meets these criteria. It can be individualized to each patient by branching to remedial and advanced sections. Furthermore, the active involvement of the user in going to other segments, and/or in answering questions, enhances the user's attention span along with his/her comprehension, retention, and interest level. The video segments also provide the opportunity for valuable active demonstrations, such as showing actual push-ups or skin inspection techniques. Topics covered in the program include:

a. causes of pressure sores
b. skin etiology
c. various ways of relieving pressure
d. how and when to perform skin inspections
e. proper use of equipment such as cushions and wheelchairs
f. who to contact and what to do if a sore developes

Through Computer Assisted Interactive Video, this information can be conveyed effectively to the patient while also freeing the therapist/educator for other activities.

CONCLUSIONS

A pressure sore clinic provides an organized, multidisciplinary approach to reducing pressure sores. By providing better seating surfaces, which minimize the buttock interface pressure, and by better educating the spinal cord injured client, the incidence of pressure sores should be reduced. Presently, the emphasis is on prescribing cushions, monitoring and evaluating clients' pressure relief profiles, and training them in the basics of pressure sore prevention. The clinic also provides a framework for transferring the knowledge gained in pressure sore prevention techniques to other centers. In addition, the expert system for cushion prescription provides the opportunity to establish a database by forming a network of centers which use the expert system and share the data collected. This should yield information on the in-field performance of different cushion types, patterns in prescription and replacement, and epidemiological data.

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THE "RESPONAUT PROGRAM" - A MODEL SYSTEM OF SERVICES AND CARE FOR LIFE-SUPPORTIVE PERSONS IN ENGLAND

10.3

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ABSTRACT

The "Responaut Program" is a hospital-based operational program which provides needed care and services for ventilator-dependent persons in England. These severely physically-disabled people are either at home with family members, or in the community, because of the development of other suitable alternatives. They are able to live, learn, and work using life supportive technology and other technical aids because of a system that they have designed with professionals and others concerned. This program began in 1965 and has served 411 former patients; as of 6/15/83, it benefits 233 people with a highly personal home maintenance service as well as the hospital base-unit (Phipps Respiratory Unit - St. Thomas' Hospital, London).

INTRODUCTION

A responaut is a ventilator-dependent, severely physically-disabled person living independently. The term was coined by the original responauts themselves; these polio survivors, like astronauts, wanted to venture out into the unknown. Responauts want to and now do live as they choose in an appropriate community setting. They now include persons with other conditions requiring life-supportive technology and other technical aids, which permit them the opportunities for learning, working, and enjoying their lives. This is made possible by a model system - the "Responaut Program".

ORGANIZATIONAL COMPONENTS OF THE "RESPONAUT PROGRAM"

The "Responaut Program" is an organized series of services and options available to anyone in England who depends upon prolonged mechanical ventilation or who has a respiratory disability that requires the Phipps Respiratory Unit (PRU).

The "Responaut Program" consists of the following services:

- comprehensive medical/rehabilitative care in the PRU;
- planning for a safe discharge to home or a variety of community alternatives;
- home care, either provided by one of 17 PRU attendant staff who live in the home of responauts, or community-based caregivers, provided by a government agency and/or governmental fund;
- home maintenance services of all respiratory technical, and personal needs on a regular and emergency basis.

The Base-Unit

The PRU remains the central focus of the "Responaut Program" because it provides the:

- security for the patients, professionals, and administrators;
- guaranteed standard of care and quality assurance;
- most coordinated operational basis;
- medico-legal basis for the protection of all involved parties.

The PRU Team

The PRU Team consists of the following team members:

- Physician: a consultant and educator;
- Nurse: an administrator and educator;
- Physical Therapists: a clinician and educator;
- Social Worker: a comprehensive care-manager;
- Home Maintenance Technician: a home care service provider.

The Home Maintenance Service

Responauts could not live at home without a dependable home maintenance service. This hospital-based coordinated service includes:

- routine maintenance of major equipment;
- emergency minor/major repair/replacement;
- resupply of small parts, inventory;
- regular observation in the home of responauts and communication to all "Responaut Program" members.

REGIONAL COMMUNITY OPTIONS AND SERVICES

Not all responauts have families to go home to. Some responauts have temporary or permanent social, educational, or vocational needs that require other options in the community setting. This is made possible by utilizing available resources and facilities which are used jointly with other members of the community. The following components will illustrate how systems can develop meeting universal needs, as well as those of ventilator-dependent persons.

Respite Care

The Netley Waterside House is a partnership between government (Borough of Lambeth), and a voluntary organization (REFRESH): a group of consumers, professionals, and concerned people. It is a purpose-built holiday home to meet multiple needs of disabled people (4 responauts), socially disadvantaged and elderly (26 residents).

Education, Care and Rehabilitation Engineering

Chailey Heritage is an established hospital, school, and rehabilitation engineering center serving 160 severely disabled children (50 day/110 residential). It has accepted and undertaken the challenge of a 14 year-old responaut.

Independent Living Center

Le Court is the original Cheshire Home. It is a community of disabled adults who share services in a group living arrangement. The responauts there have chosen this alternative for social and financial reasons.

Education, Working, Communication with Technical Aids

For all responauts living at home or elsewhere, a great deal more independence is possible because of electronic and computer technical aids designed with the input of disabled persons. Possum, Ltd. is a not-for-profit organization controlled by charities. Technical aids are bought with public funds and permit the disabled person a greater degree of physical independence and reduced need for personal care attendants.

Technical Aid and Information

The transition to home or the community requires knowledge of the availability and functioning of technical aids. The Disabled Living Foundation provides two essential services for responauts (and other disabled persons) to supplement their education and experience from the PRU.

Technical Aids Center. A permanent display of comprehensive range of devices. Product information is disseminated to health care professionals, social service workers, educators, hospital administrators, architects, designers, students, and consumers. The devices are explained by professional therapists.

Information Service. A major documentation center provides needed information for disabled persons and all those concerned. The service provides subscriptions for regular bulletins and indexed information. A registry of information responds to all requests and referrals.

Self-Help and Advocacy

The Phipps Respiratory Unit Patients Association (PRUPA) has made improvements in the PRU and made possible developments in care, service, and equipment. Its co-founder and chairperson serves all disabled persons as a member of the House of Lords and the All-Party Disablement Group. In addition, advocacy, documentation, and political action is made possible for responauts, as part of the entire disabled person community, by the Royal Association for Disability and Rehabilitation (RADAR).

Special Publications

Responaut-By, For and About Respiratory Aided and Other Gadget-Aided People is a publication describing devices, their application, and experience of responauts and other physically disabled persons.

DISCUSSION

The "Responaut Program" is a model system which meets the medical, rehabilitative, and technological needs of ventilator-dependent, severely-disabled persons. At the base-unit, the requirements for life-supportive technology are determined. Further information and assistance is provided to users of technical aids at the Disabled Living Foundation. This organization, the self-help group (PRUPA), and RADAR can provide further needed information about the availability and use of technologic devices.

Many of the mechanical, electronic, and computer devices are made with the input of the disabled consumer (PRUPA, Possum). Needed information is disseminated by the Disabled Living Foundation and RADAR. Some of this information is found in a specialized journal (Responaut).

Home care and other community living alternatives are available to responauts and to other disabled persons who need them. They are safely provided to responauts by linkages to the base-unit and the home maintenance service.

CONCLUSION

Despite the need for life-supportive technology and other technologic aids, a system can be established which enables ventilator-dependent persons a life of choice in the community. By its nature, such a system will more optimally utilize available resources and meet multiple needs of other persons with disabilities.

REFERENCE

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ACKNOWLEDGMENT

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10.4

REHABILITATION ENGINEERING TECHNOLOGY APPLIED IN A TOTAL CARE FACILITY: A FEASIBILITY STUDY

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ABSTRACT

The hypothesis of this study is that rehabilitation engineering technology applied to routine daily living activities in a total care setting could reduce medicaid costs, further the cause of deinstitutionalization, and increase independence for its residents. The results of the study took a first step toward proving that hypothesis by identifying problems and potential solutions that could be addressed through the provision of rehabilitation engineering services to the total care community. While the actual social and economic benefits have not been demonstrated, findings and conclusions of the case study at Richfield's McVitty House, a skilled care nursing facility in Salem, Virginia, are strong enough to indicate that expansion of this service to this community would meet the original intent.

INTRODUCTION

In the interests of reducing Medicaid expenditure in total care facilities, of deinstitutionalization, and of increasing the possibility for a more independent lifestyle for both the aging population and/or the more severely disabled, the Commissioner of Virginia's Department of Rehabilitative Services (DRS) authorized a study to determine the practical value of applied rehabilitation engineering technology for the total care community. To realize this, the Commissioner met with Mr. Ted Russell, President of Richfield Retirement Communities and the administrative and rehabilitation engineering staff at Woodrow Wilson Rehabilitation Center (WWRC) to propose conducting the study at Richfield's skilled care facility, McVitty House, using the WWRC rehabilitation engineering team.

The general purpose of the project was to determine if rehabilitation engineering services could be practical and beneficial both socially and economically for severely disabled and aged individuals in a total care setting. Specifically, the project team was intent on identifying generic problems in the three observation areas of bathing, mobility, and transfers which could be reduced or eliminated through rehabilitation engineering technology. Identification of more efficient methods for carrying out those daily routines would also be detailed. The team was interested in finding ways to increase the independence of the McVitty House residents and to reduce both the cost and staff time involved in total care. Furthermore, this consultation was viewed as a model experience upon which to decide the feasibility of expanding this service within the overall nursing home industry. The team, therefore, limited the scope of the assessment to three generic patient activities, looking toward identifying problems that might be characteristic of most total care facilities and developing solutions that could then be widely used. This would speak to the issue of a more long-term commitment to the expansion of such services.

METHODS

Once the cooperative agreement had been developed, the project staff formalized the implementation logistics or methodology. The final phase before implementation involved strategic planning for the three-day consultation. Who would participate in what capacity, the schedule of patients and activities to be assessed, and the format for the assessment process were completed. A female rehabilitation engineer participated to allow for the privacy of female patients who would be observed in the bathing and transfer procedures. The physical therapist from McVitty House was to observe and learn to carry-through the rehabilitation engineering function for McVitty House after the project. A physical therapist from WWRC came for one day of the three. She had prior experience in rehabilitation engineering assessments in other institutional settings and was interested in identifying problems and solutions generic to institutions.

Richfield/McVitty House staff provided the schedule of assessment activities showing resident names, any pertinent medical diagnosis, the activity to be evaluated, time of that activity and personal care attendant names. Residents were a representative sample of McVitty House in terms of age, sex, and primary physical and/or mental diagnosis. Personal care attendants were those who normally work with the selected residents in their routine activities. The schedule overall was based upon normal daily routine and activities carried out in a usual manner, employing usual techniques and adaptive equipment.

The three-day rehabilitation engineering consultation service was conducted July 25-27, 1984. The consultation was initiated with an organizational meeting to orient all key personnel involved to the assessment process and to organize, from that group, the assessment team. Included in this meeting from Richfield were the director of McVitty House, nursing staff, and the physical therapist; and, from DRS, two rehabilitation engineers, an adaptive equipment specialist, and project support staff.

The first activities assessed were bathing and related transfers. Each team observed McVitty House residents with their personal care attendants in bathing. Tasks associated with bathing and related transfers were identified, noting problems with those tasks and documenting current procedures and rationale so that when recommendations regarding the solution to noted problems were being determined, current practice could be considered.

The assessment of bathing was completed midday and the entire group - assessment team and administrators - reconvened. It was decided that more interaction with residents and their personal care attendants would be advantageous to overall findings, conclusions, and recommendations. Accordingly, the schedule was revised to lighten the patient load. Additionally, at the request of the Corporation President, assessment of restraining devices and techniques replaced the assessment of mobility on the schedule, as the current practice is considered by patients, visitors, and facility inspectors as being dehumanizing.

The second day, patient restraints were assessed. This included reviewing current devices and techniques, documenting reasons for their use, noting related problems, and suggesting solutions.

On the third and final day, the rehabilitation engineering team completed more individual evaluations. These individual assessments enabled the assessment team to determine if referral to other support services would be appropriate.

Following the on-site consultation, the assessment team met with project support staff to detail findings, identify solutions, determine the availability of the recommended solutions, and suggest potential benefits of applying those solutions. All findings and related information were then submitted to project support staff for preparation of the final report.

RESULTS

Bathing

1. Identified solutions associated with bathing are applicable to all residents, predominantly commercially available, and will result in a savings of time and energy and increased resident and staff safety.

2. Use of commercially available devices can be implemented with existing staff resources at a relatively low cost with immediate gain.

3. Implementation of all solutions has the potential to increase resident independence and decrease personal care attendant involvement.

4. Non-commercially available devices would require rehabilitation engineering design and fabrication with immediate gains relative to staff and resident time and safety and long-term gain relative to cost.

Transfers

1. Select individuals could benefit from the gradual replacement of wheelchairs without removable armrests to those with removable armrests. Use of these chairs would result in increased independence of the residents and a decrease in staff involvement. Purchase of these chairs may necessitate education of the staff in appropriate transfer techniques. These chairs are commercially available.

2. Resident and staff safety and comfort during transfers involving a gurney could be enhanced by fabrication of a gurney-to-wall latch and a one-operation brake for gurney and wheelchair wheels. These are not commercially available, but could be fabricated on-site (Richfield) with minimal cost and under the direction of a rehabilitation engineer.

Mobility

Mobility was not considered a significant issue at McVitty House as residents' food, medication, etc. are delivered by staff.

Restraining Devices and Techniques

1. Restraint methods are a significant issue at McVitty House due to the large percentage of residents requiring restraint and the dehumanizing effect perceived by residents, visitors, and facility inspectors.

2. With appropriate medical intervention, the potential exists for all restraining devices or solutions suggested to be fabricated on-site with the technical support of a rehabilitation engineer for choosing materials, determining fabrication techniques, and applying engineering theory.

3. Proper use of the restraining devices or solutions can result in increased safety for all staff and residents; easier resident management; increased functional posture; decreased debilitation; and a more humane, less restrictive environment.

Individual Residents

1. The rehabilitation engineering assessment team identified a number of solutions to client specific needs which are either commercially available or could be fabricated on-site with direction from the rehabilitation engineer. Solutions would result in increased independence, improved posture, improved self-image, improved physical status, and reduction in long-term medical complications.

2. The rehabilitation engineering assessment team identified the need for comprehensive rehabilitation services that would increase physical functioning, apply that functioning to daily living skills, and reinforce those skills through therapeutic recreational activities.

CONCLUSIONS

From the results obtained, the following conclusions can be made:

1. Rehabilitation Engineering may be able to reduce facility cost by reducing attendant involvement with residents and increasing the efficiency of routine patient care procedures, thereby saving time, energy, and money. Likewise, this service can improve the physical and emotional status of residents, thereby reducing long-term medical and maintenance/treatment costs.

2. Rehabilitation Engineering may be able to further the cause of deinstitutionalization through early intervention in the rehabilitation of aged and/or severely disabled individuals to prevent institutionalization. Through participation in screening of individuals for admission

and placement in residential treatment or care facilities, these services could prevent or delay admission to total care and/or accomplish placement in intermediate care or independent living facilities. For some individuals already institutionalized, this technology can increase independent functioning enabling discharge to an independent living situation. In all these situations, rehabilitation engineering intervention could additionally identify and bring into play other rehabilitation support services to capitalize on the benefits or gains of the applied technology, thus promoting independence and, to the greatest extent possible, self-sufficiency. This would also contribute to reduction of facility costs.

3. Rehabilitation engineering can increase the independent functioning of those who reside in intermediate and total-care communities by fabricating devices that will enable individuals to be more responsible in their daily living activities and less dependent on attendants. For these individuals, rehabilitation engineering can also improve functional posture and physical status, thereby decreasing further debilitation and reducing long-term medical complications and costs. Overall, this technology can create a less restrictive, more humane environment which will increase the comfort and improve the self-image of all concerned.

After implementation of suggested solutions, a thorough analysis of the cost effectiveness and benefits relative to solutions requiring engineering design and fabrication will be conducted.

Concurrently, a survey of other facilities may be conducted to validate the assumption that problems and potential solutions are characteristic of the overall total care community. Thus, the need for this service can be better realized.

ACKNOWLEDGEMENTS

Staff and patients at McVitty House; David F. Law, Jr., Adaptive Equipment Specialist; Rosanne Hill, Rehabilitation Engineer; and Karen Sawyer, Physical Therapist.

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10.5

COMPUTERIZED HANDLING OF INFORMATION ON TECHNICAL AIDS IN THE ITALIAN ADVICE CENTRES NETWORK AND COMPUTER-ASSISTED EVALUATION OF THE SERVICE

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FONDAZIONE PRO JUVENTUTE "DON CARLO GNOCCHI", Milano, Italy

ABSTRACT

Since 1980 the Evaluation and Information Centre on Technical Aids has been working at designing, creating, evaluating through practical use in its Advice Service and improving a computerized data bank on technical aids for the disabled.
Due to new developments, some new Advice Centres were set up all over Italy during 1984. Those centres all have direct access to the Data Bank, and form part of a developing network which will cover the whole Country. Data on the utilization of the Data Bank and of the counselling service are registered and thus statistically processed which ensures integrated field evaluation of the effectiveness of the service.

INTRODUCTION

The possibility of setting up an effective advice and information service on technical aids for the disabled in Italy started when one of the major institutions working in the field of Rehabilitation, namely the "Fondazione Pro Juventute", decided, in the frame of its research activities (Bioengineering Centre), to undertake a project aimed at creating a computerized Data Bank and an experimental Advice Centre directed at the users, the professionals, the architects and in general all the people involved in the rehabilitation process. After an experimental period (1980-1982) the project brought about the establishment of the Evaluation and Information Centre on technical Aids (SIVA) which has been offering counselling services for the last two years through its staff, its permanent exhibition and its computerized information management. The close connection between software development and practical use has allowed continuous improvement based upon actual information needs. The final version was completed by the end of 1984.
The Data Retrieval Program (designed on purpose by SIVA) is an "expert system", offering its user (to whom no experience in informatics is required) guidance in finding the information needed starting from the functional and environmental problems of each disabled person. In order to play an active role in the present project HANDYNET (aimed at creating an European information network on the technical and social resources which can promote care, rehabilitation and social integration) run by the Commission of the European Communities, the Data Bank has been given a Multilingual Access feature, allowing retrieval of data in several mother tongues even if data are stored in a different language. Technical details of the system are described in [1], [2] and [3] : but in 1984 the setting up of other Advice Centres offering counselling services in other regions of Italy was put into effect, involving major improvements in both software and organization of services. These improvements are described in this paper.

THE DEVELOPMENT OF THE ITALIAN INFORMATION NETWORK FOR THE DISABLED.

Several different technical means are offered to make available remote access to the information updated and collected by SIVA, each one suitable to the different information needs and economic possibilities of the Centres and organizations asking for connection (Fig.1). The Data Bank is hosted in a DEC VAX 11/750 Computer where updating of the contents is performed daily by SIVA's staff. A version is also hosted and monthly updated on the UNIVAC 1100/90 Computer of CILEA in Milan, which offers access through terminal and telephonic network or the data network ITAPAC. ITAPAC is in turn connected to most of the data network existing abroad.
Therefore information is accessible from any terminal in the world subject to appropriate agreements, and the multilingual access feature overcomes any problem of data translation. At the

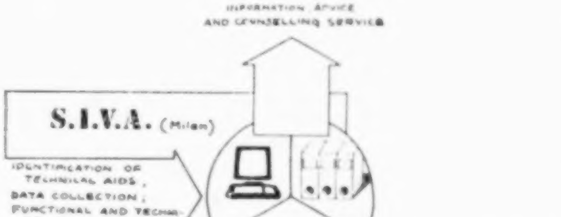

INFORMATION ADVICE AND COUNSELLING SERVICE
S.I.V.A. (Milan)
IDENTIFICATION OF TECHNICAL AIDS, DATA COLLECTION, FUNCTIONAL AND TECHNICAL EVALUATION,
computer DEC VAX 11/750
C.I.L.E.A. (Milan)
computer UNIVAC 1100/90
①
②
③
④
⑤
Rome
tape
disk
abstracts and cross reference printouts
SIP
ITAPAC
local node
Abroad (access to SIVA's information in other mother-tongues)
computer DEC VAX or DEC PDP
local files
ADVICE AND COUNSELLING SERVICE
LOCAL ADVICE CENTRE

FIG. 1

date of submission of this paper, the data bank is operating in Italian, English and French, and contains extensive information on about 3100 products belonging to some 300 different types and 500 manufacturers and suppliers. Besides SIVA's own advice centre, up to now four other centres have been established by various Organizations showing different characteristics (fig. 2):

1) the Advice and Resource Centre run by the Municipality of Turin services the rehabilitation facilities and the Local Health Authorities of the Region of Piedmont. Monthly transfer of updated magnetic tape to its DEC VAX 11/750 computer ensures connection to the information network.
2) The Cagliari Local Health Authority has set up a documentation centre servicing its rehabilitation facilities which can offer information throughout the whole region of Sardinia. Connection to SIVA is performed through periodical mailing of magnetic disk to its DEC PDP 11/34 computer.
3) The Treviso Local Health Authority's office which deals with the provision of technical aid has created a professional advice service, available to offer information in the Region of Veneto. It accesses directly the UNIVAC 1100/90 Computer of CILEA through telephone/ITAPAC network and a personal computer working as remote terminal as well as for the management of local files.
4) An association of professional and disabled, namely the "Centro Studi Prisma" in Belluno, obtains SIVA's information through periodical mailing of five updated printouts containing product information abstracts and cross-reference indexes providing basic information for choosing appropriate aids.

Each of these centres is willing to set up a permanent exhibition, the importance of which has been widely acknowledged. Contacts are being developed in order to promote the creation of centres in other Regions. Using the data bank and printouts does not require any background in but a basic knowledge of the problems of disability. However, SIVA is going to organize periodical courses for the information brokers in order to improve service effectiveness.

FIELD EVALUATION OF THE COMPUTERIZED DATA BANK.

When putting the Data Bank at the disposal of peripheral Centres, the availability of a tool able to monitor and to obtain knowledge on how the data bank is used is of paramount importance. SIVA's experience suggests that the following

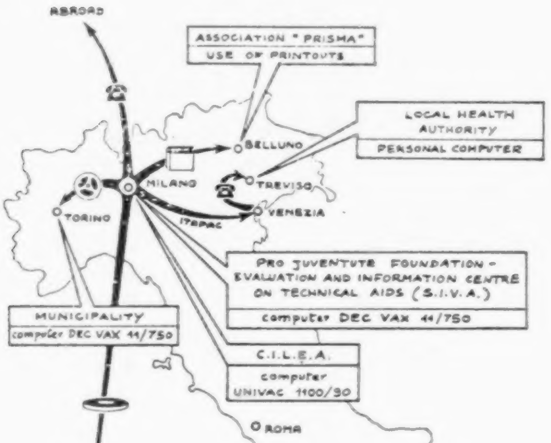

ABROAD
ASSOCIATION "PRISMA"
USE OF PRINTOUTS
LOCAL HEALTH AUTHORITY
PERSONAL COMPUTER
BELLUNO
MILANO
TREVISO
TORINO
VENEZIA
ITAPAC
PRO JUVENTUTE FOUNDATION - EVALUATION AND INFORMATION CENTRE ON TECHNICAL AIDS (S.I.V.A.)
computer DEC VAX 11/750
MUNICIPALITY
computer DEC VAX 11/750
C.I.L.E.A.
computer UNIVAC 1100/90
ROMA
LOCAL HEALTH AUTHORITY
computer DEC PDP 11/34
CAGLIARI

FIG. 2

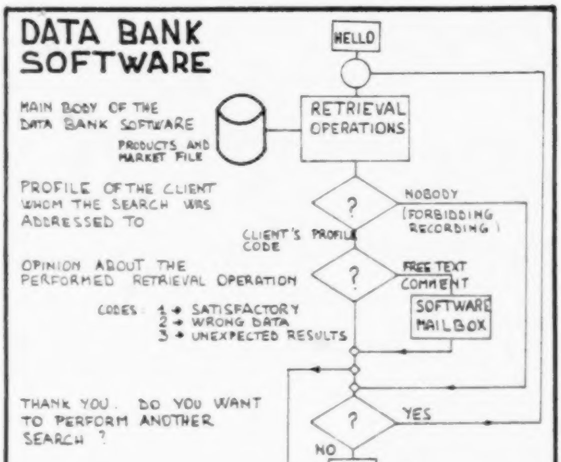
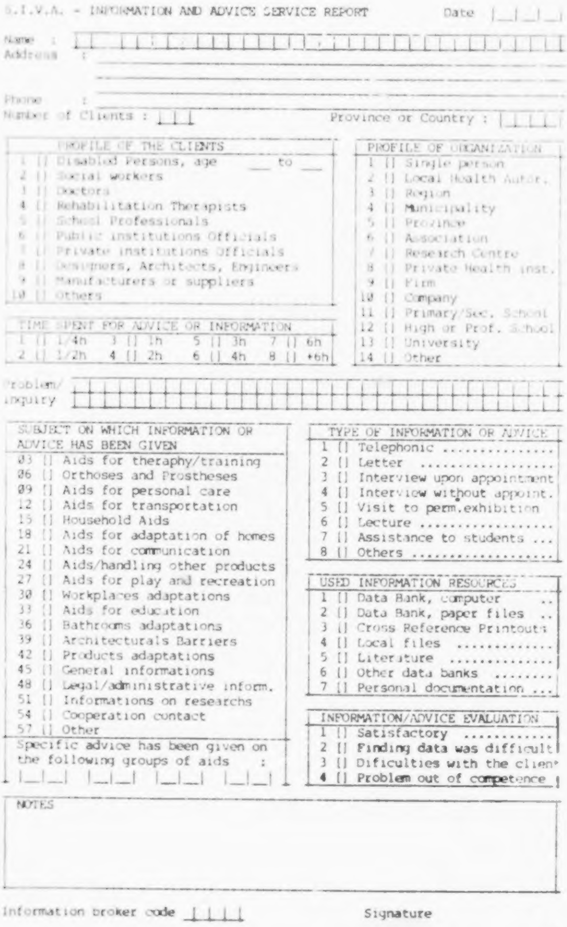

DATA BANK SOFTWARE
HELLO
MAIN BODY OF THE DATA BANK SOFTWARE
PRODUCTS AND MARKET FILE
RETRIEVAL OPERATIONS
PROFILE OF THE CLIENT WHOM THE SEARCH WAS ADDRESSED TO
?
NOBODY (FORBIDDING RECORDING)
CLIENT'S PROFILE CODE
OPINION ABOUT THE PERFORMED RETRIEVAL OPERATION
?
FREE TEXT COMMENT
SOFTWARE MAILBOX
CODES: 1 → SATISFACTORY
2 → WRONG DATA
3 → UNEXPECTED RESULTS
THANK YOU. DO YOU WANT TO PERFORM ANOTHER SEARCH ?
?
YES
NO
BYE
"HISTORICAL" FILE
DATA PROCESSING PRODUCTION OF REPORTS
ANALYTICAL CHRONOLOGICAL REPORTS
STATISTICS ON TRANSACTION AND SEARCH STRATEGIES
STATISTICS ON REQUIRED PRODUCTS AND DATA
COMPARISONS OF DATA BANK USE IN VARIOUS CENTRES
STATISTICS ON USERS' OPINIONS AND COMMENTS
DATA BANK EVALUATION SOFTWARE

FIG. 3

aspects should be emphasized:

1) knowledge of the most common transactions needed by the users, in order to improve interactive software where worthwhile;
2) identification of the data used most often, in order to devote more effort to data collection where worthwhile;
3) identification of the critical points in the interactive software (where data retrieval appears difficult or misleading and the user is likely to make mistakes);
4) getting help from the users in identifying data which can eventually be obsolete;
5) Receiving users' comments through a software mailbox;
6) Understanding when the data bank is used inappropriately, allowing SIVA's staff to contact and help the user for better management.

Through 3 simple questions the user is asked to specify, after each retrieval operation, the type of client for whom the search has been performed and the opinion of the results. If the user coope- rates, the main characteristics of the transactions are recorded and periodical analytical and statistical reports can be produced by SIVA (fig.3) allowing effective field evaluation.



S.I.V.A. - INFORMATION AND ADVICE SERVICE REPORT Date |__|__|__|

Name :
Address :

Phone :
Number of Clients :
Province or Country :

PROFILE OF THE CLIENTS
1 ☐ Disabled Persons, age ___ to ___
2 ☐ Social workers
3 ☐ Doctors
4 ☐ Rehabilitation Therapists
5 ☐ School Professionals
6 ☐ Public institutions Officials
7 ☐ Private institutions Officials
8 ☐ Designers, Architects, Engineers
9 ☐ Manufacturers or suppliers
10 ☐ Others

TIME SPENT FOR ADVICE OR INFORMATION
1 ☐ 1/4h 3 ☐ 1h 5 ☐ 3h 7 ☐ 6h
2 ☐ 1/2h 4 ☐ 2h 6 ☐ 4h 8 ☐ +6h

PROFILE OF ORGANIZATION
1 ☐ Single person
2 ☐ Local Health Autor.
3 ☐ Region
4 ☐ Municipality
5 ☐ Province
6 ☐ Association
7 ☐ Research Centre
8 ☐ Private Health inst.
9 ☐ Firm
10 ☐ Company
11 ☐ Primary/Sec. School
12 ☐ High or Prof. School
13 ☐ University
14 ☐ Other

Problem/
inquiry

SUBJECT ON WHICH INFORMATION OR ADVICE HAS BEEN GIVEN
03 ☐ Aids for theraphy/training
06 ☐ Orthoses and Prostheses
09 ☐ Aids for personal care
12 ☐ Aids for transportation
15 ☐ Household Aids
18 ☐ Aids for adaptation of homes
21 ☐ Aids for communication
24 ☐ Aids/handling other products
27 ☐ Aids for play and recreation
30 ☐ Workplaces adaptations
33 ☐ Aids for education
36 ☐ Bathrooms adaptations
39 ☐ Architecturals Barriers
42 ☐ Products adaptations
45 ☐ General informations
48 ☐ Legal/administrative inform.
51 ☐ Informations on researchs
54 ☐ Cooperation contact
57 ☐ Other

Specific advice has been given on the following groups of aids :
|__|__| |__|__| |__|__| |__|__|

TYPE OF INFORMATION OR ADVICE
1 ☐ Telephonic ....
2 ☐ Letter ....
3 ☐ Interview upon appointment
4 ☐ Interview without appoint.
5 ☐ Visit to perm.exhibition
6 ☐ Lecture ....
7 ☐ Assistance to students ...
8 ☐ Others ....

USED INFORMATION RESOURCES
1 ☐ Data Bank, computer ..
2 ☐ Data Bank, paper files ..
3 ☐ Cross Reference Printouts
4 ☐ Local files ....
5 ☐ Literature ....
6 ☐ Other data banks ....
7 ☐ Personal documentation ...

INFORMATION/ADVICE EVALUATION
1 ☐ Satisfactory ....
2 ☐ Finding data was difficult
3 ☐ Dificulties with the client
4 ☐ Problem out of competence

NOTES

Information broker code |_|_|_|_| Signature

ADVICE SERVICE MONITORING AND EVALUATION

When beginning to set up peripheral Centers a common procedure was established in order to monitor the whole service both at national and at local level and to allow information exchange and comparison among different centres. In fact, use of the data bank occurs only for a certain amount of information enquiries, which can be relevant to a broader field than product information. Each counselling service relevant to a client is recorded through a standard form (fig. 4) and stored in SIVA's computer. Software has been developed on purpose allowing production of statistical and analytical reports, which give a powerful tool for evaluation (fig.5) of all aspects of the service.

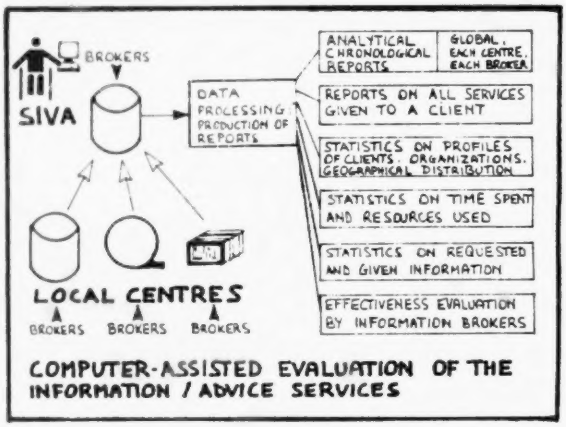

SIVA
BROKERS
DATA PROCESSING: PRODUCTION OF REPORTS
ANALYTICAL CHRONOLOGICAL REPORTS
GLOBAL, EACH CENTRE, EACH BROKER
REPORTS ON ALL SERVICES GIVEN TO A CLIENT
STATISTICS ON PROFILES OF CLIENTS. ORGANIZATIONS. GEOGRAPHICAL DISTRIBUTION
STATISTICS ON TIME SPENT AND RESOURCES USED
STATISTICS ON REQUESTED AND GIVEN INFORMATION
EFFECTIVENESS EVALUATION BY INFORMATION BROKERS
LOCAL CENTRES
BROKERS
BROKERS
BROKERS
COMPUTER-ASSISTED EVALUATION OF THE INFORMATION / ADVICE SERVICES

FIG. 5

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3) R. Andrich, S. Facchinetti, "The Italian computerized Data Bank on Technical Aids with Multilingual Access", Proceedings of the 2nd International Conference on Rehabilitation Engineering, Ottawa 1984.

ACKNOWLEDGEMENTS

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10.6

Measuring Client Gain From Rehabilitation Engineering Services

George D. McCoy, North Carolina Division of Vocational Rehabilitation Services
Frank Puckett, University of Iowa Hospitals and Clinics

ABSTRACT

Clients served by the North Carolina Division of Vocational Rehabilitation Services were evaluated to determine the effect of rehabilitation engineering services on their level of functional independence. A special assessment instrument was designed by the rehabilitation engineering staff and administered to clients served by the engineers during the period from July through November, 1983. Results of the study suggest that assessment of a client's level of independence in and around the house and in using a vehicle is a feasible outcome measure for the rehabilitation engineering program in North Carolina.

INTRODUCTION

For several years, respected professionals in the field of rehabilitation engineering have called for more studies to measure the effectiveness of rehabilitation engineering services (1,2). The approach taken in this study was to try to evaluate the impact of rehabilitation engineering services on vocational rehabilitation clients' functional independence.

BACKGROUND

Since 1979, the North Carolina Division of Vocational Rehabilitation Services (NCDVRS) has operated a field-based rehabilitation engineering service delivery program. One rehabilitation engineer is assigned to and located in each of the state's four administrative regions. With four engineers, every county in the state has access to rehabilitation engineering services. Typical engineering services provided include residence, vehicle, and job or job site modifications. The engineers serve an average of 60 to 65 VR clients per year, for an annual combined average of 240 to 260 clients. In addition, the engineers provide consultation to non-VR clients and the community as time permits. Funding for services comes from a variety of sources, such as the VR Agency, the client, Worker's Compensation, the Veterans Administration, civic or church groups, and others.

The NCDVRS decided to incorporate engineering into its service delivery system because of the Director's commitment to rehabilitation of the severely disabled. North Carolina has a statewide network of rehabilitation centers which serve as the main resource for orthopedic surgery, neurosurgery, rehabilitation nursing, physical and occupational therapy, prosthetics and orthotics, and other medical professions. Some of the major medical centers also include biomedical engineers on staff. The need in North Carolina is for professional engineering consultation in the field, especially important in a state which is predominantly rural. Although the NCDVRS' engineers may act as team members with the rehabilitation and medical center staff on occasion, they operate on their own accord as they travel to all parts of the state. These engineers primarily deal with modifying the client's environment, assuming that the client has already received maximum benefit and appropriate assistive devices from the specialists in the rehabilitation centers. In rural North Carolina, lack of accessible housing and transportation remain major obstacles to successful independent living and vocational rehabilitation outcomes. The NCDVRS rehabilitation engineering program focuses on this pragmatic level of service delivery; research does not compose an important part of the engineers' activity.

The NCDVRS' subjective assessment of its rehabilitation engineering program is that engineers help improve service delivery to severely disabled clients by increasing placement options; by providing correct design for removal of architectural barriers; by designing solutions for the driving problems of people who have severe physical disabilities; by preparing bid packages which include only the items needed to effect rehabilitation; and by monitoring the cost and the quality of work performed by vendors on behalf of clients. This subjective assessment led to the Agency's decision to convert the project, in 1981, to permanent program status at the end of the initial three year grant period. However, the Agency has continued to search for means of obtaining objective measures of the impact of the engineering service. Program evaluation is difficult because the engineering service is tailored to meet the needs of an individual client and an individual rehabilitation plan; cost effectiveness of the engineering program cannot be analyzed outside the context of overall vocational rehabilitation cost effectiveness, and control of variables in a field-based program is virtually impossible, since the clients who are receiving services are scattered across a large geographic area and do not come in to a central location for services.

Mr. Frank Puckett, then a doctoral student at Southern Illinois University, contacted the Agency because he was interested in doing a program evaluation of the North Carolina rehabilitation engineering service delivery system. He suggested using increase in a client's level of functional independence as the outcome measure. While this approach would not provide a total evaluation of the program, it seemed like a good place to start, and would help determine in an objective manner whether clients were actually benefiting from the program. Of additional worth to the Agency was that technical assistance would be available for structuring the study and for data collection; that a disinterested party would be examining the program objectively; and that some costs of the study could be supported by a grant to Mr. Puckett from the University of Wisconsin-Stout. Thus, the Agency decided to work cooperatively in this effort.

MATERIALS AND METHODS

One major aspect of this study was to develop a Rehabilitation Engineering Services Evaluation (RESE) scale, utilizing the staff of the rehabilitation engineering program in North Carolina. The specific purposes of this study were to develop the RESE scale to assess its reliability and validity, and to demonstrate its use in a study designed to evaluate the effect of rehabilitation engineering services provided to clients by the NCDVRS. Twenty-one subjects were identified from the Agency's engineering clients for the study.

In developing the RESE scale, several sources were reviewed to gain information about other scales which measure functional independence (3,4). Activities of daily living scales (5) were reviewed as well as the work done at Rancho Los Amigos on the development of their program evaluation for rehabilitation engineering services (6). For our purposes, the scale items needed to be behavioral and easily observed in order to facilitate data collection. In this respect, the ADL scales used by occupational therapists proved to be the most helpful. The following is a sample item from the RESE scale:

1. Can the client get in and out of his/her home...?
   a. Independently, with no type of assistance needed.
   b. Independently, but uses an assistive device.
   c. Partly dependent on another person.
   d. Totally dependent on another person.

Two subscales were developed for the RESE scale--residential and vehicle. Three alternative forms of the RESE scale were developed--a self-report form, a clinical assessment form, and a behavioral observation form. The self-report form was completed by each subject in the study. The clinical assessment form was completed by the rehabilitation engineer assigned to each client's case. Lastly, two trained raters interviewed each client in the home and observed his or her performance of the activities identified on the RESE scale. The behavioral observation form of the RESE scale was administered twice to each client in an effort to measure the client's level of functioning at a time prior to and following the provision of rehabilitation engineering services.

RESULTS

Ebel's statistic was used to assess the reliability across the three alternative forms of the RESE scale. The estimated reliability for the total RESE score was .80. For the RESE subscales (residential and vehicle), the reliability estimates were .77 and .76, respectively. Validity was assessed by correlating the RESE scale with other measures of functional ability. The RESE was found to correlate quite well with the other measures of activities of daily living and self-care. Correlation between the RESE scale and the Barthel Index was .79 ($p<.01$). The RESE scale was also correlated with the self-care subscale of the Life Functioning Index ($r(19)=.77$, $p<.01$).

The RESE scale has demonstrated acceptable reliability and credible evidence of validity. Most importantly, the RESE scale has shown potential for measuring the gain in level of functional independence following the provision of rehabilitation engineering services.

DISCUSSION

The Agency's experience in this study supported its subjective impression of client benefit from the engineering program. Other authors have discussed the relationship between a client's level of functional independence and potential for vocational rehabilitation and/or independent living (7,8). More research is needed on the impact of various types of rehabilitation engineering service, for example, removal of home barriers, removal of community architectural barriers, removal of transportation barriers, provision of assistive devices, development of new mobility or communication devices, etc. Further work also needs to be done to research the relationships that presumably exist between rehabilitation engineering, client functional independence, and client potential for a positive vocational and/or independent living outcome.

A problem encountered with the RESE scale which merits further research is that it was not possible to differentiate the safety or convenience aspect of performing a task quickly and easily as opposed to performing it slowly and with difficulty. For example, a client marked on the RESE scale that he could get into and out of a van independently, without assistance, because he transferred from his wheelchair to the floor of the van, dragged himself along the van floor, and pulled up into a seat. The fact that a lift allowed quicker, easier, and safer access was not detected by the RESE scale, since the client marked the scale the same way before and after the addition of a lift. Since we decided to concentrate on the most objective measures possible, we tried to avoid the more subjective nature of "convenience" in functional independence. However, this narrow focus resulted in a smaller reported increase in independence, which needs to be dealt with in future studies. This example underscores the difficulty of research on rehabilitation engineering.

Despite these problems, the Agency has found the outcome of the study to be a promising beginning for establishing a means of ongoing program evaluation for its field-based rehabilitation engineering service delivery system.

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A SURVEY OF WHEELCHAIR SEATING SERVICE DELIVERY PROGRAMS IN THE UNITED STATES

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ABSTRACT

A survey of 26 facilities in 17 states was undertaken. Information was collected on technical, patient, and financial topics. It is concluded that wheelchair seating is a sizable and widely practiced specialty. Funding and education were the main problems currently encountered by practitioners in this field.

INTRODUCTION

The field of special wheelchair seating has received a good deal of attention in the past ten years, mainly focused on technical and therapeutic aspects of postural and tissue trauma seating. New products have appeared in the marketplace, and several hundred people have received training in the provision of seating services, for example through the annual workshop hosted by the University of Tennessee. The three purposes of the current survey were: to determine the impact of the seating delivery industry in the United States; to complement previous surveys of this type (3, 4) by accumulating a greater detail of financial information; and to see what common obstacles practitioners perceived.

SURVEY DISTRIBUTION

240 questionnaires were mailed to facilities in 36 states. Responses were received from 26 centers in 17 states. The facilities were chosen by combining the mailing lists from the Seating Newsletter Club, and the attendees at the University of Tennessee annual wheelchair seating workshop.

Of the facilities replying, 12 of 26 were hospital-based, six were state-funded programs or institutions of various kinds, and eight were from private industry.

Not all 26 respondents replied to each question. In this paper, the actual number of responses to a specific question is noted.

SURVEY RESULTS

Patient Population

The 18 respondents reported providing services for 3293 patients, average 183 patients/year/facility.

Two patterns of diagnostic distribution were evident: one where Cerebral Palsy was the diagnosis of most patients; and the other where Spinal Cord Injury (SCI) or Cerebral Vascular Accident (CVA) was the leading diagnostic group.

Cerebral Palsy was the most frequently seen diagnosis in 21 of 24 facilities. It was usually followed by Myelomeningocele (12 of 21 centers) and Muscular Dystrophy (9 of 17 facilities). Cerebral Palsy accounted on average for 74% of the case load, Myelomeningocele for 13% and Muscular Dystrophy for 8%.

Two centers reported SCI, and one CVA, as the leading diagnostic category. In both cases, Cerebral Palsy was the second most frequently seen diagnosis. In these "type 2" facilities, SCI or CVA typically accounted for 55% of the caseload, and Cerebral Palsy for 38%.

Technical Aspects of Services

A total of 2822 seating systems were provided annually at the 22 centers reporting, average 128 seats/year/facility. The majority --62% -- of these were custom systems, where a custom system is defined as having a specially-contoured seat and/or back, and a standard seating system has both the seat and the back symmetric. Finally, 25 centers performed a total of 6796 repairs/modifications per year, average 272 per facility. The number of repairs/modifications exceeded the number of new seating systems at 19 of 25 facilities.

The overwhelming favorite manner of fabricating seating systems is plywood and foam constructed at the reporting facility. This was the principal method of 18 of 23 centers and accounted for 61% of the total number of units produced. Two centers used principally commercially-available plywood and foam systems, and one facility each relied mainly on Mulholland systems, on OrthoKinetics products, and on modular molded plastic inserts fabricated at the facility.

TABLE 1. TECHNIQUES FOR PROVIDING SEATING SYSTEMS

Name of Technique or Manufacturer	# of systems made this way	% of
Plywood & foam, you make	1725	61
Molded plastic, you make	484	17
Desemo/bead technique	139	5
Gilette/Chailey style	107	4
MPI	84	3
Plywood & foam, commercial	76	3
Contour U	24	--
Mulholland	19	--
Foam in Place	8	--
Britax	7	--
Other	139	5

Financial Aspects: Third Party Payors

The facilities reported that MedicAid, state Crippled Children's Services (CCS), and private insurance were the primary sources of funding for seating services.

TABLE 2. SOURCES OF PATIENT FUNDING

Source	Number of Facilities Reporting	Percent of Patients With This Funding MAX	MIN	AVG
Self pay	11	20	1	3.2
Private Insurance	16	72	5	21.2
CCS	19	95	2	23.3
MedicAid	17	97	2	28.3
MediCare	9	25	2	3.3
Service Clubs	6	40	3	3.2
Veterans Admn.	3	5	2	.5
Dept of Rehab.	6	80	3	5.7
Others	8	100	2	11.1

All three of the leading sources of funding for seating services were reported to usually require prior approval of a project. Prior approval was required by MedicAid 94% of the time, CCS required it in 88% of reported cases, and private insurance in 69% of cases.

The average time to obtain approval was reported at 33 days for private insurance, 34 days for CCS, and 51 days for MedicAid.

TABLE 3. PRIOR APPROVAL FOR SEATING SERVICES

	Prior Approval Required		Time to Obtain Prior Approval			
				DAYS		
Funding Source	# of Replies	% Yes	Replies	Max.	Min.	Avg.
Private Ins.	16	69%	6	90	1	33
CCS	16	88%	10	65	1	34
MediAid	16	94%	12	180	10	51
MediCare	5	20%	1	15	15	15
Service Clubs	9	67%	5	50	30	34
Veterans Adm.	3	67%	1	7	7	7
Dept. of Rehab.	9	89%	6	30	1	17

All funding sources were reported to pay 78 to 100% of the value of the bills submitted.

TABLE 4. RETURN ON BILLINGS

		Return or Paymt. Rec'd. Divided By Billing Rendered (Percentage)		
Funding Source	# of Replies	Max.	Min.	Avg.
Self-pay	7	100%	95%	99%
Private Ins.	9	100%	80%	89%
CCS	8	100%	84%	94%
MedicAid	10	100%	50%	87%
MediCare	5	100%	50%	78%
Service Clubs	5	100%	100%	100%
Veterans Adm.	2	100%	100%	100%
Dept. of Rehab.	5	100%	100%	100%

Facilities were asked if their home state had MedicAid codes specifically for seating systems. Apparently, five states do have these codes (SC, TN, WI, MN,and MA), and six do not (IL, VT, CA, NE, MS, and PA). From three other states (OH, MI, and KS), responses were contradictory and no conclusion could be drawn.

Financial Aspects: Facilities Operations

For professional services, 13 centers charged an average of $45/hr. Two responses differentiated between therapist ($105/hr., $96/hr.) and orthotist/engineer ($43/hr., $68/hr.) professional services. They were not included in the above average.

For technical services, 11 centers billed at an average rate of $29/hr.

Six respondents included the fee for evaluation for a seating system in the charge for the seating system. Seven facilities charged separately for the seating asessment.

Concerning profitability of the seating programs at 15 centers, there was an even split of programs tht were losing money, breaking even, and showing a profit. Nine centers reported total annual operating costs. Four were large programs ($600K, $230K, $200K, $168K), one was medium-sized ($70K), and four were small programs ($30K, $28K, $18K, $2K). Average volume was $150K/facility/year.

Of the five facilities reporting losses, three were hospital departments, one a large private foundation, and one a private rehabilitation engineering practice. Of the private industry group, five reported breaking even or better, and one (the private rehabilitation engineering practice) was losing money on the seating program.

Perceived Problems and Suggested Solutions

Survey participants perceived problems in three principal areas: concerning funding; education for different groups; and with technical issues. Main funding problems were reported as delays for prior approval, and slow reimbursement of accounts receivable. There were needs for education of staff, third-party payors, and other allied health professionals. Lack of available suitable equipment components, and high cost for same, were the principal complaints of a technical nature.

The respondents perceived most problems as suitable for cooperative solution (76%) while recognizing that some problems were ones they have to surmount on their own (24%). The most frequently suggested remedial actions were for: (FUNDING) legislation, acquisition of cost-effectiveness data, and education of third-party payors; (EDUCATION) regional training sessions, slides/brochures/audiovisual materials, and a seating journal/catalogue; and (EQUIPMENT) central cooperative purchasing of seating components.

DISCUSSION OF SURVEY FINDINGS

Impact of Seating Service Delivery Programs

The survey showed that wheelchair seating is practiced on a broad geographical basis in the United States. The value of new seating systems identified in the survey is nearly $3,000,000/year, based on a conservatively estimated average value of $1,000/system. Recalling that only 10% of questionnaires sent out were returned, and noting the evident underrepresentation of the commercially-available systems, one sees tht the wheelchair seating industry in the United States is quite large. It is a multi-million dollar industry, affecting thousands of patients annually. It is not possible to be more precise in estimating the size and impact of the seating industry from the current survey data.

Funding for Services

The survey showed that funding for seating services comes from three principal sources: MedicAid; state CCS, and private insurance. All three require prior approval for services most of the time, with an associated waiting period of over 50 days for MedicAid, 34 days for CCS, and 33 days for private insurance. Thus, most patients will wait 1 to 2 months from the time of device prescription until provision is started. Service providers can expect to be reimbursed between 87 and 94% of their submitted billings, on average, by these major funding sources. Finally, the number of states reported as having MedicAid codes for seating services was about equal to the number of states not having such codes.

Trends in Financial Operations of Facilities

Few clear trends were seen in the business practices of seating service delivery programs. An equal number of large and small programs responded to the questionnaires. An equal number of facilties were losing money, breaking even, and maintaining a profit on their seating programs. Approximately the same number of programs charged separately for seating assessments as included the evaluation in the equipment charges.

However, there was a clear hierarchy of fees for different services. Therapist services were highest-priced, followed by professional (orthotist/engineer) services, and rounded out by technical services. The average rates for professional services and technical services ($45/hr. and $29/hr., respectively) are less than comparable rates reported by Heinrichs and Shaw (3) in their 1980 survey ($50/hr. and $30/hr., respectively). Bias in samples is the likely cause.

Finally, it was apparent that the private enterprise facilities were generally more conversant with the financial aspects of their operations than were their counterparts in hospitals or state programs.

Common Problems

Respondents identified three areas of common concern: funding, education, and equipment. This is consistent with Shaw's 1983 survey (4), in which funding was perceived as the number one problem in delivering rehabilitation engineering services. A number of remedial actions were suggested in the present study. Some of these are amenable to research/development funding, such as investigation into cost-effectiveness and development of different kinds of educational materials.

CONCLUSIONS AND RECOMMENDATIONS

1) Seating service delivery should be recognized as the major field of activity it has become.
2) Facilities with diverse backgrounds and structures are working to deliver these services. Cooperation and coordination seem desirable and possible. RESNA could serve as the focal point or coordinating agency, especially on spearheading efforts on topics such as funding regulations, creation of educational materials, need for cost-effectiveness studies, and so forth.
3) Research/development funds should be forthcoming to support projects whose outcome will facilitate more widespread service delivery practices.

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COMPUTER-AIDED SURGERY

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ABSTRACT

Computer-Aided Surgery (CAS), a system for simulating the effects of surgical procedures correcting movement abnormalities, is described and compared with Computer-Aided Design (CAD), widely applied in engineering. The CAS system includes kinematic and dynamic movement measurement, musculoskeletal models, computer tomographic-based anatomic and physiological data, and computer graphics displays of anatomy and movement. The CAS system will permit simulating a surgical procedure on a computer display of the patient's anatomy and then presenting via computer animation the consequences of the putative change. By permitting iteration, evaluation and optimization of procedures before surgery, CAS promises to improve surgery performed and avoid unpromising interventions.

INTRODUCTION

Musculoskeletal disorders are reported as a major, if not dominant, source of morbidity nationally. NIH surveys ranking chronic diseases place arthritis and orthopaedic problems ahead of even heart disease. The Veterans Administration reports that, among its patient population, musculoskeletal diseases rank second in frequency of complaints, fourth in hospital admissions, and fifth in surgical procedures. Among children, much rehabilitation and surgical attention is devoted to postural and gait abnormalities. Conservative estimates put the national cost of direct care for musculoskeletal disease at $20 billion a year.

Functional assessment of the wide range of musculoskeletal disorders has become a foci of the National Institute of Handicapped Research Rehabilitation Engineering Center program and of the Rehabilitation Engineering Society of North America. Reliable means for quantitatively documenting the kinematic and dynamic state of the movement-impaired individual offers higher confidence in determining the effectiveness of surgical and/or rehabilitation procedures. This paper describes an assessment process which goes beyond documenting the status of the patient by providing simulations of the consequences of different musculoskeletal procedures affording the medical practitioner the opportunity to experiment with different approaches, to optimize the parameters of a particular process, or, decide that the simulated outcome does not warrank the hazards and liabilities of intervention.

BACKGROUND

The Newman Laboratory for Biomechanics and Human Rehabilitation at M.I.T. has been researching and developing over the past decade the major capabilities essential to a computer system to aid surgical planning for musculoskeletal disease rehabilitation and evaluate, via simulation, the changes in patient movement parameters resulting from the hypothetical surgical alteration. These include first, an accurate, reliable, rapid, and automatic system for quantifying the kinematics and dynamics of human movement; second, patient-specific, but mathematically expressed and therefore computer-manipulable, descriptions of musculoskeletal anatomy; and third, computer-generated graphical displays of both anatomy and movement, presented so as to be familiar to, and interpretable by, the medical practitioner.

COMPUTER-AIDED DESIGN

Before outlining in more detail these necessary capabilities, it is useful to discuss the computer-aided design (CAD) routinely employed by engineers in the design and evaluation of new products and processes in all competitive commercial, industrial, aerospace, and military settings where CAD is complemented by computer-aided manufacturing (CAM). In fact, computer-aided surgery is the logical sequelae to computer-aided design.

The power of CAD/CAM was evident during its origins at M.I.T. in the 1960's (1). CAD synergistically links the unique human attributes of goal identification, innovation, and cognitive association with the unbeatable speed, memory, and reliability of the digital computer. The engineering designer operates in his familiar and productive media--drawings; the computer redefines the picture into its language--bits. Once in the computer, the picture can be manipulated geometrically and can be analyzed via mathematics and the laws of physical science, with interim results presented by the computer to the designer for his consideration, and most probably, iterative change. Thus CAD combines the human's cognitive capabilities and the computer's manipulative power.

Observation of surgeons and surgical practice in orthopaedics suggests that the engineering designer and the orthopaedic practitioner have much in common. They observe the circumstances of the situation and devise an idea for a solution. Whereas the engineer-designer now can carry out the exploration, iteration, and optimization of that design in consort with the computer (after which he instructs the CAM system to program the numerically controlled machine tools and robots to make the part), the surgeon is constrained to a single solution, the particular surgical procedure performed in the operating room, and then must await the recovery of the patient to observe the consequences.

We believe a computer-aided surgery system can do for surgery what computer-aided design has done for engineering.

MOBILITY ANALYSIS

We have developed, demonstrated, and technology-transferred a unique human movement data acquisition, quantification, and display system which meets the following requirements:

- . A normal and natural milieu for the human subject, minimizing artificial aspects of the experimental environment and the burden on the subject.
- . High precision, three-dimensional, kinematic data, with body-segment translations and rotations relative to a laboratory fixed reference frame, at high data frame rates relative to the frequency components of human movement, in a form suitable for subsequent dynamic analyses.
- . Automaticity--no human intervention in data acquisition and quantization to eliminate human subjectivity and error and to reduce drudgery.
- . Very rapid processing of kinematic and forceplate data to provide kinematic and dynamic data in real time--e.g., to provide access to results during or immediately after movement.

Our TRACK software complements the Selspot cameras and Kistler forceplates to process the kinematic data while our NEWTON program uses the kinematic data together with forceplate data and body-segment inertial properties to estimate the net forces and moments across the skeletal joints (2).

MUSCULOSKELETAL ANATOMY

Patient-specific anatomical data are essential for, one, accurate determination of the mass properties used in Newton's equation to calculate forces from acceleration; second, specializing mathematically expressed musculoskeletal models to the particular patient's anatomy; and third, providing to the surgeon via computer graphics terminals realistic and accurate displays of relevant areas and volumes of the patient's anatomy.

We have demonstrated the practicability of using computer tomography data to calculate automatically the inertial tensor, by converting CT number and grey scale into local density (3). We are developing software algorithms to extract automatically from CT data, and store efficiently in computer memory the geometrical information necessary to generate colorgraphic computer displays of patient anatomy, i.e., the skeletal bones, joints, muscles, etc. (4).

Musculoskeletal modelling has been a focus of Newman Lab research for some time. Accurate kinematics and dynamics make feasible the estimation of the time course and force level in the redundant participating muscles producing a movement pattern (5)(6); however, the mathematically expressed models which capture the three-dimensional geometry and physiology of the skeleton, joints, muscles, and ligaments of, i.e., human lower extremity, should also be adjusted to the parameters, i.e., bone dimensions, muscle/tendon origins and insertions, of the particular patient. Our CT and more recently Nuclear Magnetic Resonance (NMR) data scans include the information necessary for such individualization of our musculoskeletal model.

Thus we have been aggregating the competence to document and display the patient's anatomy and his movement together with his skeletal and muscle forces during movement. With this quantitatively expressed data, we can mount a computer-aided surgical design process.

AN EXAMPLE OF COMPUTER-AIDED SURGERY (CAS)

The future competence of the CAS system is suggested by a hypothetical scenario: A patient presents with a gait abnormality. In addition to the surgeon's personal observation of the human's movement (and all of the examinations and tests he now conducts), the patient's movement pattern is acquired and recorded by mobility instrumentation (1), e.g., the three-dimensional positions and orientations of all relevant body segments at a high enough frequency to capture important dynamics, quantization of the foot-floor interaction, and synchronized myoelectric signals from the musculature. This computer-stored record goes into the patient file, but is also immediately displayed to the surgeon so that he can correlate his personal observation of the patient's movement patterns with computer-reconstructed animation and graphic display of the acquired data.

An inverse Newtonian analysis combines the experimental kinematics and forces with CT or NMR determined body-segment mass and inertial properties (3), and the computer predicts the forces and moments across joint articulations which produced the observed movement patterns (5). The computer estimates of force and moments across joints are subsequently partitioned into the temporal activity and force levels generated by the individual muscles anatomically deployed across the joints of interest (6).

Our generalized musculoskeletal computer model would be morphologically specialized to this particular patient using the CT or NMR data. Thus it would be possible to present to the surgeon on a graphics interface an anatomically correct, properly scaled picture of the parts of the anatomy on which he might chose to practice surgery (4).

Suppose, for example, an osteotomy is suggested to relieve the symtoms of osteoarthritis, removing a wedged-shape section of bone from the femur to realign that bone and femoral head with the acetabulum cup in the pelvis, thereby bringing into articulation more normal cartilage. With the computer light-pen serving as scapel, the surgeon (like the designer) could assess the proper size and orientation of wedge, "sever" the femur and "remove" the bone fragment. Then, under command from the surgeon, the computer would reconstruct the bone (as nature would do over a much longer period of time), and the computer would play back through the musculoskeletal model and into the animated walking figure display, via computer graphics, the changes in gait dictated by the choice of and site of the osteotomy.

With all of this happening with very minimal delay, the surgeon then observes the consequence of his surgical decision, and if he decides (as the engineer-designer frequently does) that his first choice was not an optimal one, he changes his procedure with the ease of erasing the screen and starting anew with the intact femur. After sufficient iterations of this simulated procedure produces an acceptable gait pattern in the animation, the surgeon is prepared to perform the actual operation. Perhaps an additional next step, by analogy to computer-aided manufacture, would be devising more precise means of transposing the computer's memory of the optimum osteotomy to the surgeon's procedure on real bone.

The premier criterion of this computer-aided system, as was the case two decades ago in our research on computer-aided design, is the determination to capitalize on, and in no way diminish, the knowledge and skills of the human operator--in this case the surgeon.

Parts of this CAS system are already in place; other parts are under development. The linking up of the entire scheme and demonstration with surgical practitioners is an ambitious undertaking. Given the remarkable utility and cost-effectiveness of computer-aided engineering design, the success of computer-aided surgery appears foreordained.

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FACTORS AFFECTING RETURN TO WORK
IN UNEMPLOYED LOW BACK PAIN SUFFERERS

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ABSTRACT

As part of an ongoing prospective study, 73 currently unemployed subjects with low back pain as a primary impairment were given a battery of tests in order to assess which factors would be predictive of return to work six months later. The subjects were drawn from either a local low back pain clinic or from the regional pool of Social Security Administration disability benefit claimants. Several factors which bear a statistically significant relationship to return to work are presented.

INTRODUCTION

Low back pain (LBP) has been estimated to affect 80% of the population sometime in their adult lives and is a major cause of disability. Despite the great impact of LBP on national compensation payments and lost earnings, there is too little known about the determinants of occupational disability due to LBP. More precise assessment of such factors would enable an improved prediction of return to work capability, which would, in turn, allow better determination of degree of impairment for claimants of disability compensation. It would also help those involved in rehabilitation of persons with LBP to select those persons most likely to benefit and to monitor their progress toward rehabilitation.

MATERIALS AND METHODS

As part of an ongoing prospective study, 73 currently unemployed subjects with low back pain (LBP) as their only major impairment were given a battery of tests in order to assess which factors would be predictive of return to work. The battery of tests, questionnaires and interviews were given within one day and included the following:

General background information;
Medical history and physical examination;
Motion and stress in previous employment;
Pain severity and effect on life;
Medication usage;
Biomechanical tests of lifting, strength, and range of motion;
Psychiatric interview;
Minnesota Multiphasic Personality Inventory (MMPI).

The total of 73 subjects comprises one group of 54 subjects recruited from the patients of a local spine clinic and another group of 19 who were claimants for Social Security Administration disability benefits (but who had not yet received a decision on benefit status). The clinic patients were selected for this study if they were between the ages of 15 and 55, had no more than one back surgery (not less than six months previously), had worked for at least three months prior to their unemployment, and had been out of work no longer than 18 months. The Social Security claimants were required to meet only the criterion of LBP being the primary impairment and, therefore, might be expected to be at greater risk of not returning to work.

All subjects were followed up six months after testing by telephone to determine their unemployment status during the intervening time. The response variable considered in this report is employment status (yes or no) at six months after testing and is designated as "return to work."

RESULTS

Various factors were correlated with the subjects' return to work status, with the following results. (The numbers in parentheses are the subject counts for the respective groups. All factor relations reported were statistically significant at the .05 level.)

1. 50% of the clinic group (54) had returned to work while only 16% of the SSA group (19) had returned.
2. Only 28% of those subjects having leg pain associated with their LBP (40) returned to work while 58% of those with no associated leg pain (33) had returned.
3. Only 28% of those having definite radiculopathy with their LBP (18) returned to work while 58% of those without radiculopathy (33) had returned.
4. Only 13% of those who can rarely distract themselves from their LBP (15) returned to work while 48% of those who could sometimes distract themselves from their pain (58) had returned.
5. The longer the subject had been out of work at the time of testing, the less likely was his return to work. For example, 53% of those out of work less than six months (43) returned while only 23% of those out of work more than six months (30) returned.
6. Only 10% of the subjects who felt that it was not very easy for them to change occupations (20) actually returned to work while 53% of those feeling it was either very or somewhat easy to change occupations

(53) did return to work.

7. Only 24% of those subjects who had not been sleeping well lately (29) returned to work, while 52% of those sleeping moderately or very well lately (44) did.

8. 58% of the subjects rated by a clinician as having high potential for return to work (36) did return while only 8% of those rated unlikely to return to work (13) did. This rating was based on a one-hour semi-structured psychiatric interview.

9. Only 33% of those subjects who have ever been smokers (54) returned to work, while 65% of the nonsmokers (17) returned to work.

Some additional factors of marginal statistical significance ($p < .10$) are also summarized below:

1. None of those with claudicatory leg pain (5) had returned to work while 44% of the others (68) had returned.
2. Those subjects returning to work rated their pain lower the day after testing than did those not returning to work. The pain was measured as a distance on a Visual Analog Scale.
3. Those subjects returning to work had worked more hours per day on average (9.3) in their previous employment than had those not returning to work (8.4).
4. Those subjects returning to work had worked more years in their last job on the average (9.2) than had those not returning to work (5.9).
5. Those subjects returning to work had had to push or pull less weight by themselves on the average in their previous employment (165 pounds) than had those not returning to work (287 pounds).

DISCUSSION

Numerous factors were considered regarding their potential relationship to return-to-work status at six months after testing and only factors having relations statistically significant at the 10% level or lower were presented above. The analyses presented here are of a bivariate nature and do not consider conditioning effects of other variables on these relationships. Also, the analyses were performed on data from the entire group of tested subjects and did not consider relationships within separate clusters of subjects on any basis. Multivariate analyses will be performed as part of the ongoing prospective study.

Just as interesting as the variables which bear some statistically significant relation to return-to-work are the many other factors which did not have any simple bivariate relationship. These include standard biomechanical tests measuring range of motion in flexion and extension, strength in flexion and extension, and lifting ability, as well as various psychological factors (e.g., MMPI scores) and other aspects of previous employment.

ACKNOWLEDGMENTS

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AN INSTRUMENT TO MEASURE RANGE OF MOTION

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ABSTRACT

An instrument to quantify range of motion has been developed as part of a computerized test battery for assessing human sensory and motor functions. It consists of a mechanical arm with four segments joined together by instrumented hinge-type joints. The tip of the arm can be easily moved to any point on the surface of a standing or sitting person. The LSI 11/23 minicomputer first calculates the xyz-coordinates of the tip and then determines either range of motion parameters or limb segment lengths or anatomical contours from sequential positioning of the tip. Test-retest reliability for range of motion assessment about the elbow joint in 10 young subjects was excellent (r>0.90) and values compared favorably with those reported elsewhere.

INTRODUCTION

Quantitative measurement of joint range of motion is important for physical therapists, orthopedic surgeons, and rehabilitation engineers. It is the foundation for prescribing meaningful therapeutic exercises, for monitoring the progress of such exercise, designing orthoses and aids, and knowing when to modify or terminate the treatment program. It plays an important role in the design of a rehabilitation strategy for handicapped individuals. It also plays a major role toward achieving an overall "assessment profile", developed recently [1] to quantitatively assess sensory and motor functions of normal and handicapped individuals.

The most commonly used instrument to determine joint range of motion is a universal goniometer. A skilled technician or therapist can use a universal goniometer quite accurately and reliably. Evaluations of the measurements are commonly carried out manually. Often, goniometers are instrumented to provide electrical signals that are proportional to the angles measured. However, the practicality of goniometry as a routine clinical or rehabilitative procedure is limited. Factors that apply to pocket, instrumented goniometers, or both can include: 1) standard devices do not exist; 2) different goniometers or special attachments are required for each joint that is tested; 3) each device must be attached to body parts on either side of a tested joint and thereby require considerable time for complete evaluations; 4) soft tissue at attachment points leads to slippage or variability in measurements; and 5) results must be hand recorded. In pilot or specific research studies these disadvantages can be justified. However, test administration time constraints alone would prohibit inclusion of such goniometry in the broad computer-automated battery of sensory and motor function tests under development [2]. Therefore, a new range of motion instrument was developed.

METHODS

Design.

The range of motion instrument was designed to form an integral part of a broad battery of sensory and motor function tests. It not only measures joint range of motion, but also limb segment lengths and it can trace anatomical contours. It is easy to operate and relatively small in size. As shown in Figure 1, the mechanical arm of the instrument is made of four hollow aluminum tubes of 19 mm diameter that are connected by tight tolerance ball bearing joints. The tolerance of the base bearing is adjustable which eliminates virtually any mechanical play in this important joint. The segments of the mechanical arm vary in length from 15 to 61 cm. The axis of rotation of the base joint is perpendicular to the mounting surface (such as a table top), the remaining three axes of rotation are parallel to the table top. This particular orientation of joints significantly simplifies mathematical computations without compromising resolution or the handling of the instrument.

Each joint voltage is calibrated by adjusting offset and gain potentiometers in the amplifier

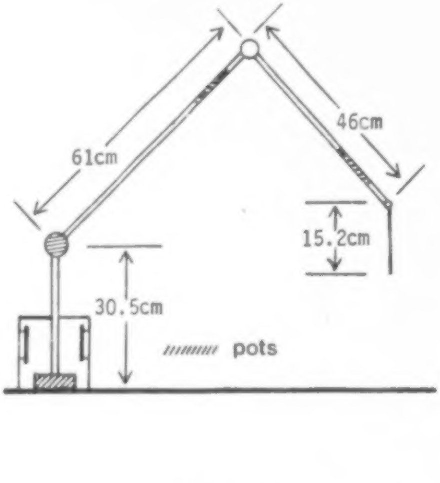

Figure 1
Design of the Range of Motion Instrument

section of the device. Output voltages of about -9 V and +9 V (d.c.) are produced respectively when the joints are rotated from one limit to the other. All d.c. supply voltages to the potentiometers are stabilized to ensure long term measurement reproducibility. A two-pole analog lowpass filter keeps noise at a minimum. The analog voltages are digitized and stored in the LSI 11/23 computer.

Mathematical manipulations transform the spherical coordinates given by the joint potentiometers to cartesian coordinates, compute angles and lengths and reduce the digital noise in the system by averaging repetitive samples of each xyz-point. The sine and cosine functions necessary for these operations are obtained from a 512 point lookup table to minimize execution speed. Then all calculations are reduced to the four basic operations of addition, subtraction, multiplication and division and, therefore, can be directly transferred to a microcomputer.

When clamped to the table top, the tip of the instrument can be held in the hand of the technician (like a pencil) and be moved easily to any point on the surface of the patients body, who is standing or sitting a few feet away from the table. After several xyz-coordinates of the tip have been obtained corresponding to the coordinates of specified anatomical locations, assessment parameters such as the range of motion of a specific joint are computed.

Bench Test.
Before the instrument was evaluated in the intended clinical application, it was bench tested. The instrument is calibrated by first positioning all joints into one extreme position and then into the other. Calibration constants are stored in the program and are checked periodically. Reproducibility was tested by keeping the tip fixed in one position and continuously measuring the xyz-coordinates. Reproducibility of the readings is (0.8+0.5) mm. This is the cumulative effect of all errors due to the 12 bit A/D conversion, the limited resolution of the joint potentiometers and the calculation errors from the lookup table. Accuracy was tested by measuring the length of a 30 cm ruler in various orientations and positions of the 1x1x2 m 3D space available to the instrument. The average resolution lies between 1 mm and 4 mm. It progressively increases as the ruler is placed further away from the base joint. This error cannot be avoided when using spherical coordinates, since the resolution of the base potentiometer is limited in the present device.

Range of Motion Evaluation.
To evaluate the device in a typical application, an experiment was carried out to measure range of motion of the elbow joint. Nine healthy male individuals, between the ages of twenty and forty, volunteered. They all declared themselves to be right-handed. They were tested twice on each arm with several minutes between the tests and with the start of each test from a complete repositioning of the subject. Subjects were seated during the tests with the upper arm parallel to the torso. The angle between complete extension and flexion is determined from the inner product between the two vectors. Each vectors is determined by measuring two points on the ulnar bone: one close to the elbow and the other one close to the wrist. Two such measurements are made. The first measurement is taken during complete extension, the second one during complete flexion. Subjects are instructed to neither rotate the upper arm nor sway the body when the four points are obtained. Such movements would skew the results.

RESULTS

Test-retest results are shown in Table 1. A linear correlation exists between the two readings with a correlation coefficient of 0.952 for the right arm and of 0.903 for the left arm. The standard errors of the linear correlation estimates are 1.819 and 3.471 degrees, respectively. These results relate very well to the data given by Moore [3], particularly when one considers that the operator was completely untrained in range of motion measurements.

Table 1

Test and Retest Results from Elbow Joint
Range of Motion in Degrees

right arm		left arm	
test	retest	test	retest
129.1	131.1	134.8	136.2
137.0	142.4	140.0	139.0
139.8	142.5	141.1	137.5
140.3	139.7	143.6	147.2
141.3	142.0	145.0	139.1
145.9	145.9	145.9	147.4
146.3	146.9	148.7	152.2
146.4	145.5	153.4	150.5
146.4	145.5	158.8	157.7

DISCUSSION

We are pleased with the results obtained and consider them to be encouraging for future clinical applications when the device is fully integrated into the computer-automated assessment system. We expect to be able to reduce the instrument error discovered during tests by using infinite resolution potentiometers. However, for range of motion applications the current accuracy appears to be sufficient.

A new methodology is introduced for range of motion measurement. In conventional goniometry, the therapist aligns the universal goniometer with the estimated midline of the upper and lower arm of the patient. He or she attempts to eliminate variations in the measurements due to soft tissue structures as much as possible. An experienced therapist has a variance of three degrees in any two measurement trials [3]. It is difficult to duplicate the universal goniometer technique with the present range of motion instrument, because it is based on the measurement of individual points instead of projecting two lines on top of each other. Therefore, differences arising from soft tissue movements during the rotation of a limb are not easily eliminated. Patients also vary considerably in size, shape and anatomical structure. We therefore changed the technique for measuring range of motion slightly, to that presented above.

While the simple range of motion application is presented, a primary reason for selecting the specific design is to facilitate measurements of anatomical contours. Such measurements are not possible with goniometers. Commercially available devices could be used as a range of motion instrument and an anatomical curve tracer [4,5]. Their cost, however, is considerably higher than ours without a significant gain in accuracy, resolution or ease of handling. In Table 2 an example is given of the xyz-coordinates (in cm) of a human spine that was obtained with our instrument. The instrument was placed on top of a 74 cm high table. The axes of the cartesian coordinate system do not coincide with the sagittal and frontal planes, which is obvious when the numbers of Table 2 are plotted. However, it is theoretically possible to align the axes of the coordinate system in parallel to the base by mathematical manipulations.

TABLE 2

XYZ-coordinates of a Normal Human Spine

x	y	z
97.0	-9.3	59.4
94.9	-9.3	55.9
91.6	-9.2	51.1
88.6	-8.7	46.7
85.7	-9.1	42.0
84.8	-8.7	39.0
84.6	-8.7	33.5
84.8	-8.9	30.4
85.6	-9.3	26.9
87.1	-9.7	22.3
87.9	-10.5	18.5
89.3	-11.4	11.5
89.0	-11.8	6.2

This data is presented here to demonstrate how spacial coordinates of anatomical structures can be entered into the computer. The data can then be further manipulated to extract parameters that have clinical significance, such as the radius of curvature and the spacial orientation of certain regions of the spine. As progress is made toward full development of such techniques, we expect the instrument to become an integral part of the computer-automated sensory and motor function laboratory.

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4. 3SPACE Digitizer and Tracker, Polhemus Navigation Sciences Division, McDonnell Douglas Electronics Company, PO Box 560, Colchester, VT 05446, (802) 655-3159.

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11.4

QUANTITATIVE FUNCTIONAL ASSESSMENT AND CHARACTERIZATION OF HEAD INJURED PATIENTS

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ABSTRACT

Fifteen patients at various stages post head injury were evaluated quantitatively with a computer-automated sensory and motor function test battery. The purpose of the study was to first assess and then, if findings were encouraging, use results to demonstrate the applicability of the test battery to this population. Furthermore, we desired to determine strategies for future evaluations, and to begin functional characterization of the head injured population during rehabilitation. Results, presented in terms of functional profiles, indicate that the test battery offers potential to provide accurate documentation of rehabilitation progress in this disabled population.

INTRODUCTION

A computer-automated sensory and motor test battery has been under development for the last five years (1). The system currently consists of instrumented devices integrated into a computerized data acquisition system and a data base management system. Quantitative assessments of upper and lower extremity strength, speed, coordination, range of motion, reactions, and sensations at various body sites, as well as hand/arm tremor, body stability, mental status, and activities of daily living can be quickly administered by a trained technician. Most tests have been evaluated for test-retest reliability and a large normal population has been evaluated (over 300 subjects, ages 20-80 years) to establish reference functional levels, the utility of which are discussed elsewhere (2). We have planned studies with this broad assessment system (300 different measures of function are envisioned, of which approximately 200 can now be obtained) to determine applicability to various handicapped populations and to determine device or system modifications to enhance use. Here, we introduce the methods used in one such study involving a head injured population.

BACKGROUND

Head injury often leads to functional problems in cognitive, psychologic (emotional), and physical (sensory and motor) areas. Evaluation of the latter has been in terms of general or gross categories such as mobility, and often only end treatment function has been reported (3,5). Aside from dexterity assessment with the Purdue pegboard, detailed physical function evaluations from date of stabilization after injury through the end of rehabilitation has not been commonplace. A recent review of the literature revealed several studies of head injured patient function (3-9). All of the cited studies rely on coded rating scales, which have sensitivity and reliability limitations (10). Standardized rating scales have yet to become accepted, perhaps because of these limitations.

METHODS

Fifteen head injury patients, 12 males and 3 females, were tested on a selected subset of tests in the computer-automated system. Age ranged from 17-39 years with a mean age of 26.67 years. Time from injury onset to test date ranged from 3-27 months with a mean of 10.07 months. With the exception of one subject, all tests were completed in one session per subject. The same examiner, a trained occupational therapist, administered all tests. The 62 measures of function obtained are listed in Table 1 by descriptive name and are grouped by functional category (dominant and nondominant side are considered separate measures). Details of each test and administration procedures are given elsewhere (1). Each test listed lasts approximately 5 to 20 seconds. As with all subjects evaluated, demographic information was obtained at the start of the session and was recorded with test results on each subject's floppy disk.

Test results were first processed to form a data record, a procedure that involves processes such as averaging the best 2 of 3 trials of a particular test. Each data record was then entered into the sensory and motor function data base management system. With special features in the data base management system, each patient's data record was compared to a normal population of the same gender and age decade to produce results for each test in terms of the number of standard deviation units from the normal population mean. For most tests, our normal data base in this age range is fairly robust, with more than 50 observations for each measure. To facilitate presentation of results in the limited space available, the simple average (no weighting factors) of all measures within a function category was computed to form a category composite result.

Table 1. Measures of sensory and motor function obtained from 15 head injured patients.

VISION
  Central visual efficiency, R and L eye

MEMORY
  Short-term

ACTIVITIES OF DAILY LIVING
  Putting on a shirt
  Zipping a zipper
  Tying a bow
  Buttoning large button
  Buttoning small button
  Manipulating safety pins
  Threading a needle

MANUAL DEXTERITY
  Large peg manipulation, D and ND
  Purdue pegboard, D and ND

UPPER EXTREMITY STRENGTH
  Grip, D and ND
  Wrist dorsiflexion, D and ND
  Extended arm abduction, D and ND

UPPER EXTREMITY REACTIONS
  Simple visual reaction time, D and ND hand
  2 choice reaction time, D and ND hand
  8 choice reaction time, D and ND hand
  Arm sweep reaction time, D and ND

UPPER EXTREMITY SPEED
  Index finger tapping, D and ND
  Hand-arm tapping, D and ND
  Lateral alternating reaching, D and ND
  2 choice movement, D and ND
  8 choice movement, D and ND
  Arm sweep, D and ND

UPPER EXTREMITY COORDINATION
  Lateral reaching and tapping - D and ND
  Hand-eye random target tracking - D and ND (4 measures each)

LOWER EXTREMITY STRENGTH
  Foot dorsiflexion - D and ND
  Extended leg flexion - D and ND

LOWER EXTREMITY REACTIONS
  Simple visual reaction time, D and ND foot

LOWER EXTREMITY SPEED
  Foot tapping, D and ND
  Lateral reaching leg swing, D and ND

LOWER EXTREMITY COORDINATION
  Lateral reaching and tapping - D and ND

Note: D - dominant side,
ND - nondominant side

RESULTS

Results are presented as function profiles as shown in Figures 1, 2, and 3. Figure 1 represents the average and range for the 15 patient population in each of twelve major function categories. The least disability is indicated for vision, memory and upper and lower extremity strength and speed. Figure 2 shows a similar plot for a patient 3 months post injury, showing functional below that of the group in most categories, while Figure 3 illustrates the functional profile for a patient 27 months post injury.

DISCUSSION

Results are not presented to make a major statement about function in head injury patients, but rather to demonstrate the possibilities for undertaking studies that may do so. We conclude that tests in the computer-automated battery can be administered to the head injured population and that functional profiles of the type presented can be useful to track patient progress and document rehabilitation trends. We can observe in Figure 1 that physical function is extremely variable in this population, as expected. While individual results are presented for two patients, one recently injured and the other injured more than two years prior to the test date, no attempt was made to normalize results based on severity of the initial injury. Unfortunately, data for the second patient at 3 months post injury was unavailable for comparison. Therefore, while the plots demonstrate the expected trend (the patient further from injury date has better function), there were exceptions to this trend in our data set. We found such exceptions to be strongly correlated with the nature and severity of the initial injury, as evidenced by other medical information available.

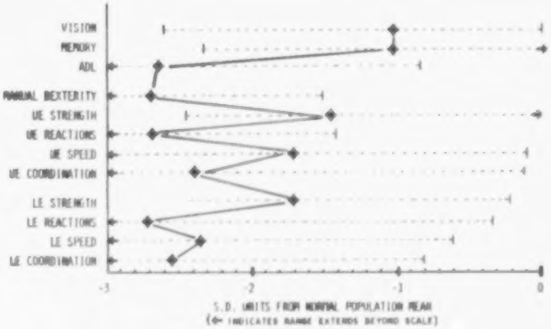

Figure 1

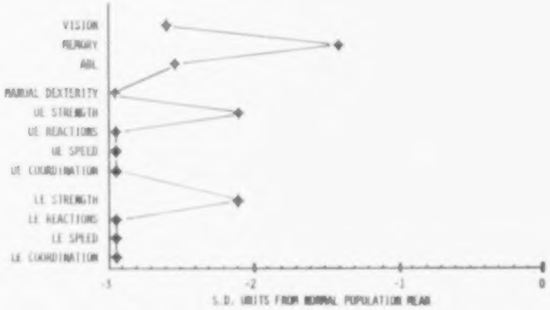

Figure 2

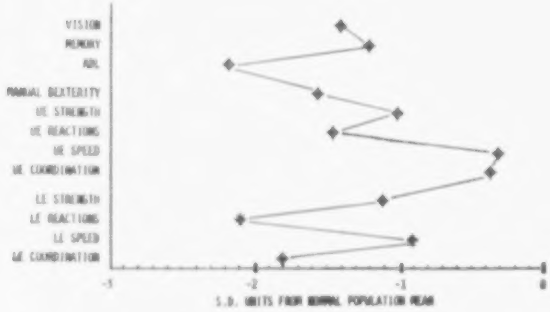

Figure 3

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It should be noted that most tests were administered to right and left body sides. In the above results, right and left side results were averaged, but could easily be displayed as separate profile points. Future studies are planned to determine the effect of pre-injury dominance and post injury dominance switch on head injury rehabilitation outcome.

With a variety of head injury recovery trends possible, as well as a choice of rehabilitation methods, the availability of a broad scope quantitative test battery to evaluate these trends and treatment modalities effectively is presented. Questions such as, "When should therapy (or a portion of a therapy program) be discontinued?" could be answered based on hard data. The system also provides the capability of objectively documenting changes in function for rehabilitation reimbursement purposes, important for treatment of other populations in addition to the head injured.

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11.5

Graphical Sensory and Motor Function Profiles Based on Computer-Automated Quantitative Assessment

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ABSTRACT

We have developed a computer-automated test battery to quantitate a broad range of sensory and motor functions and an associated data base management system. This system is viewed as the front end of an overall system to facilitate objective and quantitative assessment of function for a variety of applications. Methods and software packages that have been developed to present interpretable, application specific functional profiles are presented and expert system concepts are discussed.

INTRODUCTION

While much research is carried out routinely to improve the function of handicapped and impaired individuals, methods for objective and quantitative functional assessment have not advanced significantly in a coordinated and directed way over the years. Ordinal rating scales, requiring the skilled but subjective observation and judgments of experienced clinicians, are still considered to be state-of-the-art assessment instruments by many. Decisions concerning comparative efficacies of therapies and devices often rely on these techniques.

We have developed a computer-automated battery of instrumented devices to objectively and quantitatively document sensory and motor function (1). The computer-automated system includes assessments of mental alertness, vision, hearing, steadiness, reactions, tactile sensations, manual dexterity, speed and coordination, posture, selected activities of daily living, strength, and fatigue. Multiple measures are obtained for each of these general categories, depending on number of sites evaluated and the number of measures required to characterize a particular function.

Until recently, our primary focus has been on the development of test devices, administration procedures, and the data acquisition end of the process. However, in order to evaluate a patient's sensory and motor function on the basis of measurements collected with the newly developed computer-based system, there is an associated need to establish standardized and simplified methods of interpretation. Here, we describe our latest progress toward this end, as well as anticipated next steps in the development process.

BACKGROUND

Over the last several years, the prototype computer-automated test battery has been refined and improved, new tests (range of motion, proprioception, pronation/-supination, speech, respiration, pain, resistance to passive motion, and gait) and a sensory and motor function data base management system have been or are being developed, the test battery is being evaluated clinically, and studies to demonstrate applications of the system are being carried out. The ability to collect more than 300 different measures selectively is envisioned.

With the present system, data is collected with physical units (milliseconds, kilograms, bits per second, etc.) that are appropriate to describe and quantitate a given function. Numerous difficult decisions that affect data management were made along the way (2). Difficulty is encountered when a clinician is presented with a printout of these raw test results and is expected to interpret them. These difficulties arise from several factors: (1) there may be more than 150 measures of function on a single printout; (2) there are many different physical units to comprehend; (3) in order to identify positive and negative findings, the clinician must have some knowledge of normal results for each different test; (4) it is known that normal function changes with age and is different for males and females, and thus normal results must be mentally adjusted by the clinician to take into account the subject's age and gender. These factors make meaningful interpretation of raw test results essentially impossible by all but those clinicians that are intimately familiar with the test battery. Therefore, in order to put the system into wider practical use, it is necessary to develop a software system that will allow clinicians to assess a patient objectively, accurately, and easily. Such a system should relieve the clinician's burden of memorizing a host of normal test result values and present results in a format that is immediately interpretable.

As an important step to facilitate this process, a sensory and motor function data base system has been designed. It is emphasized that the prime purpose of this system is to facilitate interpretation of test results and the conduction of systematic studies based on the data base, as opposed to mere mass storage of data.

The framework of the automated data base management system, with interactive test result inspection capabilities, has been implemented. At present, the data base (VAX 11/780 based) contains more than 700 records of sensory and motor function, of which approximately one-half represents patients with various handicaps. The other half represents normal data that is essential to interpret patient findings properly. It is now possible, through a totally computerized process, to collect any or all of approximately 150 measures of sensory and motor function, deposit results in the data base, and examine a patient's results by comparison to a selected subset of the normal population. Interactive utilization of the available command set is accomplished via dial up modem so that remote terminals at any site (such as our laboratories and clinician offices) can access results.

While we once recognized this as an end goal in itself, we have begun to view this as only the "front end" of an overall system. The rest of the system is conceived to consist of specialized software packages to facilitate interpretation of data. Different packages, specific to applications or types of interpreters (vocational therapists, neurologists, occupational therapists, physical therapists, orthopedic surgeons), are planned. These packages utilize a common input, namely a patient's test session data record with results normalized to the appropriate gender and age. This data is then combined, restructured, or further processed according to procedures or rules routinely used by various disciplines and then displayed in a format designed for the application. Thus, the process constitutes the essential components of specialized expert systems to allow interpretation of test results in the context of the task at hand or a given user.

COLOR GRAPHICS FUNCTION PROFILES

With the now widespread use of personal computers and their increasing color graphics capabilities, we decided to take advantage of this power to extend the data base system utility. In this arrangement, a Texas Instruments Professional Computer replaces a standard terminal at a remote site. Connection is still via a dial-up modem link. Software was developed to transfer normalized data records from the VAX 11/780 to the PC memory for further processing.

As an example of the different possibilities, a color graphics presentation of function profiles was developed. In this presentation, each body part is represented as a commonly used "pie" display. Several of these pies are arranged so as to form a close correspondence to a human body viewed in the standard anatomical position (Figure 1). Each pie is subdivided into a number of segments, each of which corresponds to a specific measure (such as pronation strength) obtained for that body part (Refer to Figure 2 list for the arm for example). Some artistic license is taken in display design to account for whole body measures such as body sway, and certain activities of daily living. Each color in an 8 color spectrum is designated to represent a defined range of function (for example, red=-3 to -2.5 standard deviation units from the normal population mean, and green=-0.5 to 0.0 standard deviation units) The pie segments are colored according to data record contents in these units (functional level). Thus, the patient's overall function can be quickly evaluated by inspection.

Display at this level, however, does not allow labeling of each pie segment for identification of measures because of screen resolution limits. Only experienced users would be able to recognize these assignments. Thus, facility is provided to allow the user to move a screen cursor on top of any given pie and press a key that generates a magnified and appropriately labeled view of the selected pie. This allows a top down approach to examination of the patient.

It is emphasized that decisions that lead to display generation are based on objective and quantitative test results compared to an appropriate normal population.

The above process represents only simple structuring of results for special displays. Often, the clinician is interested in positive findings. We have therefore developed a processing algorithm that allows the interactive user to select a threshold (in S.D. units on the normal population scale) and have only a two color display generated. Segments representing a function below the specified threshold (positive findings) are colored red, while those suprathreshold are colored green.

Several displays for trend analysis have also been designed. The most basic also uses the two color display and a specified change threshold. The user selects two test dates, normalized data records from each test date are downloaded to the personal computer, and the difference over time is computed for each measure (each pie segment). Measures showing improvement greater than the specified change threshold appear green and others are displayed as red.

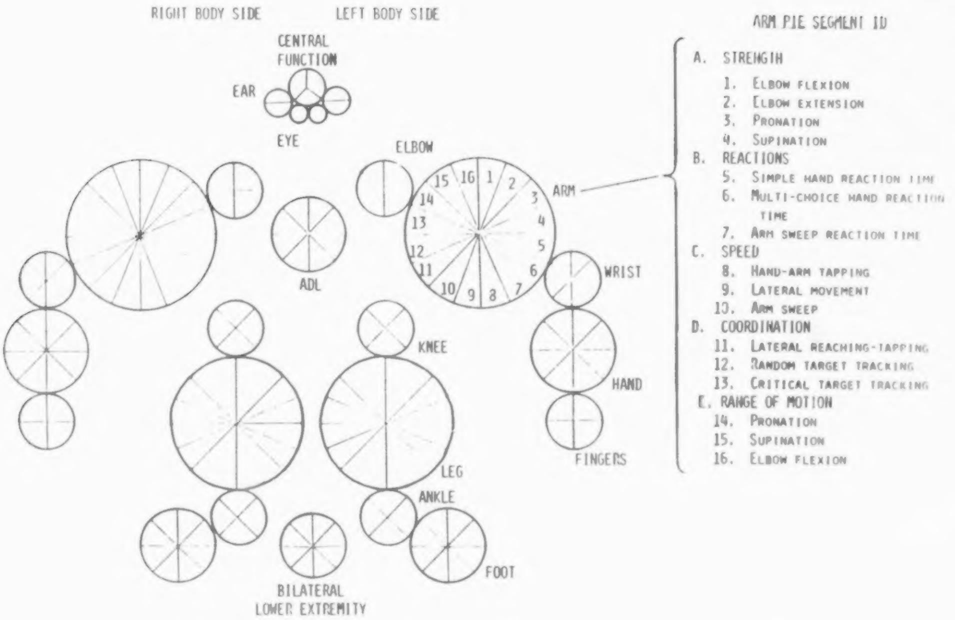
RIGHT BODY SIDE
LEFT BODY SIDE
CENTRAL FUNCTION
EAR
EYE
ELBOW
ARM
ADL
WRIST
KNEE
HAND
LEG
FINGERS
ANKLE
FOOT
BILATERAL LOWER EXTREMITY
1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
ARM PIE SEGMENT ID
A. STRENGTH
1. Elbow flexion
2. Elbow extension
3. Pronation
4. Supination
B. REACTIONS
5. Simple hand reaction time
6. Multi-choice hand reaction time
7. Arm sweep reaction time
C. SPEED
8. Hand-arm tapping
9. Lateral movement
10. Arm sweep
D. COORDINATION
11. Lateral reaching-tapping
12. Random target tracking
13. Critical target tracking
E. RANGE OF MOTION
14. Pronation
15. Supination
16. Elbow flexion

Figure 1. Color graphics display format for presentation of sensory and motor function test results.

DISCUSSION AND CONCLUSIONS

With the approach described, it is possible to develop special application specific software packages that are based on the personal computer. Such packages would have various levels of sophistication. We are planning a package to process test results and interpret them to provide results in terms more meaningful to a neurologist, with classifications such as cerebellar disorder, pyramidal tract lesion, etc. In addition, a hierarchal processing approach is also planned. With this approach, more basic functions are first examined before a display of other results is generated. As an extreme example, display of hand-eye coordination test results would be meaningless if the subject exhibited poor vision. With appropriate studies, it is theoretically possible to create a package for vocational assessment, whereby the contents of a data record representing a comprehensive functional level test session are matched to a list of possible job skills.

We believe that the methods described can become powerful assessment tools. The approach offers the potential to bridge the gap between the detail of instrumented quantitative assessment techniques and the needs of the clinician. Since the entire process from collection of individual measures to generation of final printed reports and displays is computer-automated, significant progress toward standardization of assessment should be realized.

ACKNOWLEDGMENTS

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11.6

A CLINICAL TEST FOR QUANTIFYING FOREARM PRONATION AND SUPINATION FUNCTION

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ABSTRACT

We present a device and methodology for quantification of forearm supination and pronation. During two 10 second computer-automated tests, measures of range of motion, isometric strength, and velocity of supination and pronation are obtained. The device was field tested with a group of 17 normal subjects (9 tested twice) and 20 patients with either spinal cord injury, strokes, head injury, multiple sclerosis, or low back pain. Results of these tests are presented to document the characteristics of test measures and to demonstrate force velocity of the tests. Test-retest reliability in the normal population was found to be good (r>0.80) for strength, speed, and total range of motion measures.

INTRODUCTION

Hand dexterity depends in part on the ability to supinate and pronate the forearm with speed, range, and force. Forearm contractures, disturbances in reciprocal agonist and antagonist innervation (dysdiadochokinesia), pronator syndrome, total elbow replacement surgeries, and other conditions (PNI, cerebral palsy, hemiplegia, quadreplegia) require quantification of disability severity and documentation of change with medical intervention.

The test described was developed to be integrated into a broad computerized battery of objective, quantitative sensory and motion function tests (1,2). Since tests in this system are administered as a battery, test administration time and instrumentation costs were considered to be significant design constraints. Because true quantification was desired, coded rating scales were not considered because of limitations that have been attributed to them (3). It is in this context that the proposed methods are presented. Data used in tabulating results represents part of a study of greater magnitude to characterize different patient populations by function.

BACKGROUND

Pronation-supination of the forearm occurs between the radius and ulna. The radius rotates in an arc around the ulna during pronation-supination. Movement at the distal end of the ulna during this motion is complex, but can be considered translational compared to the circular motion of the radius. There is minimal movement of the humero-ulna during pronation-supination.

Several investigators have studied forearm movement using cadavers (4), and less invasive goniometric investigations (5,6,7). Youm et al (4) demonstrated in cadaver specimens that passive forearm supination and pronation range of motion averaged 85.0 +4 and 70.0 +5 degrees, respectively, with respect to the standard neutral position. In 1982, Okada and Okada (7) reported a method for quantifying alternating forearm movement that used a minimicrocomputer and an instrumented Diadochometer. Two handles mounted to a fixed base were attached to each of the subject's hands with a bandage, and as the subject pronated and supinated, the angular position of each handle was measured. Velocity, duration, range, maximal slope, and regularity between and within intervals (smoothness) could be deduced from these measurements. It is noted that range of motion angles can have an error introduced because the hand is attached to the measuring device and direct observation of the radius and ulna is not obtained. Such errors are considered to be small, however. While providing quantitative results, this device was developed to assess a specific type of dysfunction and results are not presented in a manner easily used by the clinician. Strength assessment was not included. Because commercial devices are unavailable, most reported clinical studies (8,9,10) have relied on more crude coded rating scales and pocket goniometers to determine intervention strategy outcomes.

METHODS

Device. A device was developed that mounts to one of the assessment clinic tables, with a handle that can be grasped easily by the subject. The handle is attached to an instrumented shaft with an axis of rotation in the plane of the table. A subject is positioned facing the device with the elbow flexed to 90 degrees and hand grasping the handle. For administration of the first part of the test, the subject is instructed to rotate the handle from a neutral position, (with handle grasp vertical), and rotate it alternately to the right and left, "as far and as fast as possible" for a 10 second period. During this time, the computer monitors the angular position of the handle. After a 10 degree change in angular position is detected, the computer begins timing the test and continues to monitor angular position to compute the maximum clockwise and counterclockwise excursions in real time for each cycle, as well as the time required for each excursion. The end of the test is signaled by a computer-generated tone, and final results are calculated and displayed. Software is structured to identify and store only the locations and time of occurrence of peaks during this motion. This avoids the need for large memory blocks and off-line signal processing. The hand used (right or left), entered by the technician prior to the test, determines whether clockwise or counterclockwise rotations correspond to pronation or supination. The measures obtained during this 10 second dynamic test are average velocity (separately for pronation and supination), total range of motion and bias angle (average midpoint of range) from the neutral position (positive if in pronation, negative if in supination). Standard pronation and supination angles can be computed since the total is the sum of pronation and supination and the bias is the difference.

For isometric strength tests, a pin is inserted into the top of the device to fix the handle in the neutral position. This also engages a load cell that is calibrated to measure torque applied to the handle. During the test, the subject generates a clockwise or counterclockwise maximal torque over a 3 to 5 second trial. The peak torque is recorded by the computer as a measure of strength for separate pronation and supination tests. As with other strength tests in our battery, results are normalized by dividing by body weight. Thus, results have units of Kg-m/Kg. To obtain units consistent with other tests (% body weight), the moment arm (2.85 cm) between the rotation axis and load cell is used to convert torque to force. We realize that this makes results dependent on device geometry, but the procedure facilitates interpretation of results and device geometry is known and fixed.

Evaluation. Normal adult males (N = 8) and females (N = 9) were tested once. Nine subjects were retested no less than one and no more than two weeks later. In addition, several patient groups were evaluated when the device was introduced into our computer-automated battery of tests. They included spinal cord injury (N = 7), stroke (N = 8), head injury (N = 6), low back pain (N = 1), and multiple sclerosis (N = 1). While it is intended to evaluate these subjects periodically during their rehabilitation therapies, present results represent a snapshot in time of this long-term process. Therefore, statistics presented for patient groups represent baseline data. However, data was extracted from our data base for one patient repeatedly evaluated to demonstrate trend documentation.

RESULTS

Results for normal subjects are shown in Table 1, while Table 2 summarizes results of patient tests by group. Change in function over time for a 40 year old female stroke patient is shown in Table 3.

Group normal statistics indicate that strength of males is greater than of females, even when results are normalized and expressed per unit body weight. Speed measures are similar to reported values (7), but total range is approximately 10 degrees greater than expected. In addition, results show pronation angles to be greater than supination, contrary to reported values (11). Except for the bias angle measure, test-retest correlation coefficients were good. Most patient groups (spinal cord injury, head injury, stroke, and multiple sclerosis) showed functional deficits. The low back pain subject (male) showed functional levels comparable to the male normal group.

Test-retest evaluation of the single stroke patient indicates progress during rehabilitation, with gains in strength and total range of motion. However, increased speed is documented only for the nondominant forearm.

TABLE 1. RESULTS OF NORMAL PRONATION AND SUPINATION FUNCTION MOVEMENT

Measurements	All Normals						
	Base Line		Test-Retest	Males Only		Females Only	
	Mean	S.D.	r	Mean	S.D.	Mean	S.D.
STRENGTH (% body weight)							
Pronation, D	51.1	18.1	0.83	60.9	19.1	44.2	15.8
Pronation, ND	49.8	17.9	0.81	57.8	20.1	43.9	14.0
Supination, D	53.2	12.5	0.92	59.7	13.2	48.6	10.0
Supination, ND	51.3	15.2	0.87	52.9	19.6	43.1	19.8
SPEED (degrees per second)							
Pronation, D	772.1	330.2	0.86	882.7	289.4	729.2	370.1
Pronation, ND	717.9	262.5	0.84	804.5	240.8	644.6	266.6
Supination, D	936.0	334.5	0.85	1112.7	389.2	797.1	207.1
Supination, ND	958.7	281.5	0.83	951.0	317.9	964.2	264.8
RANGE OF MOTION (degrees)							
Total, D	171.2	41.7	0.95	179.2	39.5	165.5	43.6
Total, ND	162.2	45.5	0.85	158.6	40.8	165.0	50.2
Bias, D	0.0	23.0	0.71	+ 3.9	25.0	- 2.8	21.9
Bias, ND	- 6.7	26.9	0.69	+10.3	28.3	+ 1.8	25.7
Pronation, D	85.6	20.8	-	93.5	19.8	79.9	21.6
Pronation, ND	87.8	22.7	-	89.3	20.5	84.3	25.3
Supination, D	85.6	20.9	-	85.7	19.7	85.6	22.0
Supination, ND	78.4	22.8	-	69.0	20.3	80.7	24.9

Note: D - Dominant forearm, ND - Nondominant forearm

TABLE 2. RESULTS OF PATIENT PRONATION AND SUPINATION FUNCTION MEASUREMENT BY GROUP MEANS

MEASUREMENTS	SPINAL CORD INJURY	STROKE	HEAD INJURY	MULTIPLE SCLEROSIS	LOW BACK PAIN
STRENGTH (% body weight)					
Pronation, D	34.2	31.0	40.9	22.7	60.8
Pronation, ND	39.3	18.0	37.4	31.8	50.8
Supination, D	33.5	28.5	41.7	35.4	61.4
Supination, ND	39.7	20.6	29.0	33.7	41.6
SPEED (degrees per second)					
Pronation, D	284.3	307.8	463.3	-	1120.0
Pronation, ND	215.7	167.1	395.0	-	1070.0
Supination, D	312.9	453.3	780.0	-	850.4
Supination, ND	280.0	208.6	593.3	-	650.3
RANGE OF MOTION (degrees)					
Total, D	117.1	136.9	139.2	-	187.0
Total, ND	88.0	81.3	92.5	-	157.0
Bias, D	+25.3	+27.2	+20.0	-	+ 5.0
Bias, ND	+18.7	+ 0.6	- 3.8	-	+ 3.0

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TABLE 3. CHANGE IN FUNCTION FOR 40 YEAR OLD FEMALE STROKE PATIENT

MEASUREMENTS	CHANGE	
	Absolute Percent	S.D. Units on normal female population scale[a]
STRENGTH		
Pronation, D	+ 6.4	+ 0.12
Pronation, ND	+ 3.6	+ 0.08
Supination, D	+ 38.5	+ 0.55
Supination, ND	+ 16.7	+ 0.17
SPEED		
Pronation, D	- 12.5	- 0.08
Pronation, ND	+ 22.2	+ 0.08
Supination, D	0.0	0.00
Supination, ND	+100.0	+ 0.53
RANGE OF MOTION		
Total, D	+ 18.0	+ 0.41
Total, ND	+ 14.0	+ 0.12
Bias, D	+ 5.8	+ 0.14
Bias, ND	- 2.3	- 0.23

[a] (Test 2 - Test 1)/(Standard deviation of normal female population from Table 1)

DISCUSSION

Results indicate expected trends (dominant side better than nondominant side, patients showing dysfunction), which lend face validity to the proposed device and also demonstrate its utility in assessing a broad range of disabled population. Test-retest reliability in the limited number of normal subjects, indicates that repeatable measurements sensitive to small but significant differences, can be obtained. We intend to repeat these calculations as sample sizes increase.

The larger than expected total range, pronation angles greater than supination, and poor test-retest correlations for bias angle indicated a problem. Upon further investigation, it was found tht subjects were not routinely instructed to grasp the handle firmly throughout the test. The handle inertia during rapid movement would allow extra pronation range (toward loosely flexed fingers). We have since added a strap to guarantee tight coupling and modified instructions to the test subject with noticeable improvement in results. With these corrective actions, we conclude that the method can provide quick and accurate assessment of pronation and supination function.

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DYNAMIC POSITIONAL AND ELECTROMYOGRAPHIC MONITORING OF SITTING POSTURE

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ABSTRACT

This study was designed to examine the activity of selected muscles (back extensors, hamstrings) as well as trunk deviations from midline of both normal children and children with cerebral palsy (5-8 years of age) in response to changes of the seat inclination. Data revealed that muscle activity profiles of C.P. children were clearly unlike those of normally developed children. Inclining the seat forward facilitated spinal elongation and a more upright posture.

INTRODUCTION

Postural abnormalities of joints and the contractures of the surrounding muscles are recognized as important complications in cerebral palsy. The spinal column is often heavily implicated in such postural deformities. Malalignment of the spine can produce dysfunctions varying from acute and chronic back disorders due to disc or spinal nerve involvement, to severe impairment of ambulation/ mobility, sitting posture or respiratory functions in persons with the most severe spinal deformities. In addition, an upright head posture and consequent proper visual distances and perceptions are reliant upon a proper spinal alignment.

Conservative treatment of actual or potential spinal deformities must commence as soon as the child is diagnosed as potentially at risk. Adaptive seating for moderately to severely involved children is thought to be useful in controlling spinal curvatures that are flexible (Trefler, Tooms and Hobson, 1978). However, there are at present no conclusive studies that prove this to be true. At present, there is a hiatus of objective data in the literature related to the efficacy of conservative intervention, that is, physical therapy and seating equipment in the treatment of spinal curvature in the mild to moderately involved child with cerebral palsy. One reason for this paucity of information is the lack of objective assessment tools. The research described in this proposal represents an attempt to expand upon the small existing body of knowledge related to dynamically monitoring spinal posture in cerebral palsy.

BACKGROUND

The pathogenesis of scoliosis in cerebral palsy is complex and muscle imbalance and the presence of primitive reflexes are suspected of playing a role. It also appears that gravity affects the severity of the scoliotic curve with the curve increasing when individuals are placed in the sitting position (Madigan & Wallace, 1981).

Although no exact incidence figures were found for kyphosis, it too is recognized as a deforming spinal posture, usually of a less debilitating nature than scoliosis. Thoracic kyphosis can develop in ambulatory patients who have excessive lumbar lordosis consequent on hip flexor (iliopsoas) or quadriceps contractures. The lordosis is a compensatory mechanism to offset the anteriorly tilted pelvic and consequent forward shifting of the centre of gravity. The resultant thoracic curvature is an attempt to offset the lordosis and once more move the center of gravity into a more functional position. In addition, protracted shoulders and a concomitance of high thoracic kyphosis are frequently observed in spastic patients, particularly in those who use crutches.

Biomechanical and physiological studies show that the trunk in sitting is maintained upright in a condition approaching an unstable equilibrium by the continued interplay of various muscles correcting any sway away from the position of balance, in other words, by the expenditure of energy, through low grade trunk muscle activity (Floyd & Roberts, 1969). It is unlikely that such a situation exists in the sitting position in children with cerebral palsy where automatism of muscle action as well as spatial orientation are disturbed by neurological impairment. Imbalance in muscle tonus creates multiple variables such as extensor thrust, adductor contracture, hamstring tightness, anterior or posterior pelvic tilt which should be observed by long-term dynamic monitoring. There is a strong correlation that exists between the degree and rapidity of development of a spinal deformity and the imbalance of activity of agonists and antagonists around the hip joint (Rang, Douglas, Bennet & Koreska, 1981). Unilateral hip dislocation in the non-ambulatory patient can lead to pelvic obliquity which in turn, can cause seating problems, pressure sores on the weight-bearing ischium, and secondary scoliosis.

Seating and positioning devices are an essential component of the management of children who spend a great deal of the day in a sitting position, and the major concerns of therapists are to decrease the influence of abnormal reflexes by proper positioning, facilitating of coordinated interactions of muscles and the promotion of stability, particularly in the trunk and pelvis. Questions are constantly asked in relation to efficacy of adaptive seating in providing trunk fixation and in the control of reduction of spinal deformities.

There has been controversy in the literature over whether seats should be inclined forward or backwards. Bendix & Biering-Sorensen (1983), in their literature review found that some investigators recommended a 5° posterior tilt of seat angle for normal subjects. Nwaobi, Brubaker, Cusick & Sussman (1983), showed that a posterior seat angle does not decrease extensor tone in cerebral palsy. In addition, posterior tipping away from the work surface is also not feasible for these children. Another body of opinion focused on the possible advantages of anterior tipping of the pelvis and the consequent maintainance of a lumbar lordosis, decrease in hamstring tension and proper positioning for desk work. (Bendix et. al., 1983)

A basic ergonomic principle is that static muscular work should be reduced to a minimum. This principle is true for both standing and sitting, but because of the effects of gravity some static work is required to maintain a desired body position (Andersson & Ortengren, 1975). It is with this principle in mind that this project addresses the seating needs of the mild to moderately involved child with spastic cerebral palsy who sits in school in an unsupported sitting position. These children are at risk for spinal curvatures. The study has been designed to address the following issues: 1) the effect of an anteriorly tipped seat on spinal posture; and 2) the effect of a 10° or a 15° anterior seat tilt. It is also anticipated that the EMG data related to the back muscles will eventually be valuable in treatment planning and evaluation.

METHODOLOGY

To assess the structural effects of altering seat base angles on the body, a tracking system was developed which monitors a point on the body - in this case, the top of the head - in three dimensional space. This device is a refinement of a single-dimension tracking system presented earlier (Bablich, Tetleborn, Sochaniwskyj & Koheil, 1984). This new mechanism incorporates two lightweight 38 cm rods, connected in series by three potentiometers. Placing a voltage across the potentiometers and measuring the change in voltage as tracking point was moved, the change in angles of the linkages relative to known, fixed coordinates could be determined. (Sochaniwskyj, Bablich & Koheil, 1985).

Eight channels of information were sampled by a Tektronix 4052 desktop computer. These channels were: five channels of EMG, and the three potentiometer voltages for the coordinate calculations. Data collection of each of the processed and integrated EMG signals is at 50Hz, and of each of the potentiometer voltages, at 5Hz to ensure an accurate representation of the signals of concern.

Bilateral paraspinal electromyographic potentials were recorded simultaneously at the mid-thoracic and lumbar levels, and from the hamstrings to provide information concerning action, fatigue and relative strength while the seat base was at various angles.

EMG signals were monitored and preprocessed by an Autogen 1700 Myograph. The 100-200 Hz bandpass of the EMG was selected for all the muscles, with the raw signals being full-wave rectified, and then integrated over 50 ms before sampling.

SUBJECTS

Ten children, 5 normal (control group) and 5 with spastic diplegia (independent sitters) aged 5-8 years of age are participating in the study. The Ss were well motivated and do not exhibit intellectual retardation.

DESIGN

The study consisted of 4 sessions each being divided into two ten-minute phases. A 20 minute period of monitoring begins to approach the functional sitting time for these children during classroom activity. Figure 1 graphically depicts the design of this study and seat angles are varied as indicated.

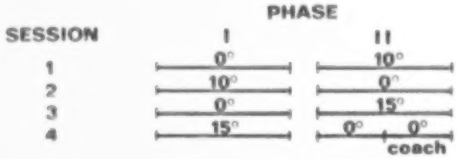

FIGURE 1: Representation of the research design indicating seat base inclinations during both phases of monitoring for each of the four sessions. During the final five minutes of the fourth session, the subject will be instructed to "sit up as straight as possible" (coaching).

RESULTS

Figure 2 presents an example of a 20-minute assessment session with an 8-year-old non-neurologically impaired child recorded while the seat inclination was 0° during the first phase, and 15° during the second. Generally, tipping the seat forward resulted in a more upright posture combined with lower mid-thoracic spinal muscle activity, although lumbar EMG levels were increased. Essencial findings from the 10 subjects are as follows:

1. EMG data agrees with findings in the literature relating to EMG studies in scoliosis i.e. maximal output is on the convex side of curve and lesser output on the concave side of the curve.
2. Alterations in EMG levels of spinal musculature, in response to an altered seat angle are observable.
3. Deviations from midline sitting posture can be determined.
4. Although the number of subjects is small, it appears that an anteriorly tipped seat facilitates spinal elongation and maintenance of midline position in the cerebral palsy child with a dynamic spinal curvature.

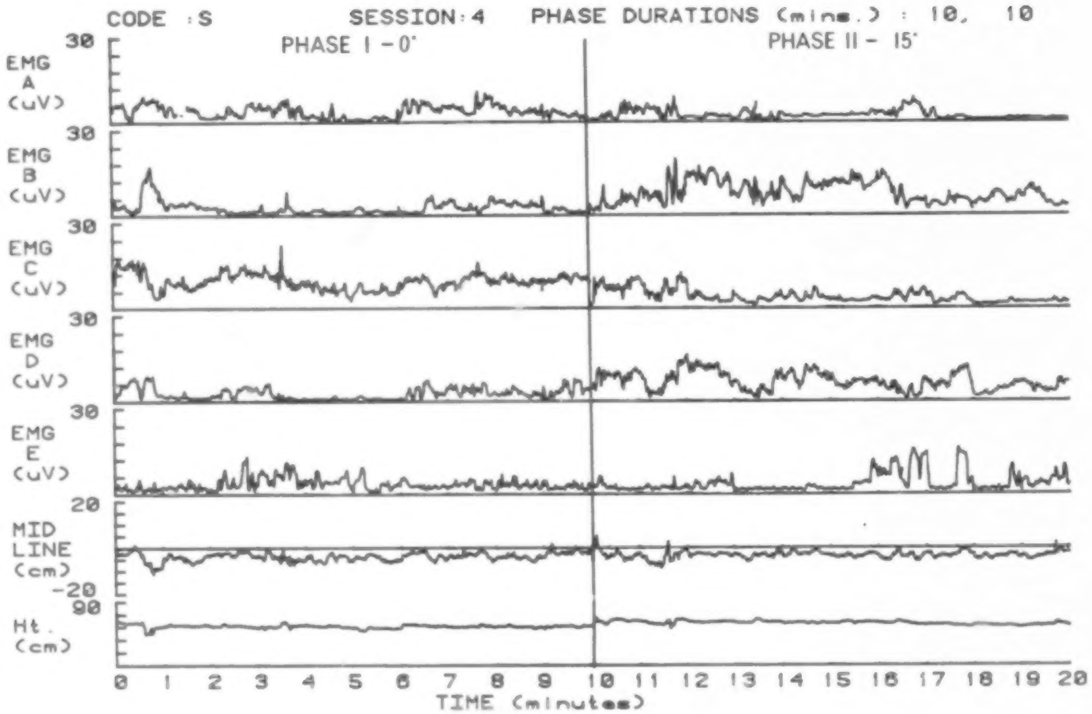

FIGURE 2: An example of a 20-minute assessment session of a normal subject consisting of two 10-minute phases. During phase I, the subject sat on a 0° seat inclination, and during phase II, on a seat with a 15° forward inclination. The five muscles are: (A) left mid-thoracic, (B) left lumbar, (C) right mid-thoracic, (D) right lumbar, and (E) hamstrings. 'Midline' refers to the subject's deviation from a midline reference point, and 'Ht' refers to the subject's vertical height measured between the seat base and top of the head.

DISCUSSION

The combined procedure of EMG monitoring of spinal musculature and 3 dimensional point tracking gives a comprehensive picture of spinal posture and midline deviation of the trunk during sitting. The effects of seat angle inclination on sitting posture can now be monitored over clinically relevant periods of time. This work will ultimately be important in: providing objective data in relation to sitting posture; determining an ultimate seat inclination to effect optimal spinal posture; assessing treatment interventions, and alterations in seating equipment. This monitoring procedure would be applicable to both paediatric and adult populations.

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11.8

RESPIRATORY INDUCTANCE PLETHYSMOGRAPHY IN CHILDREN WITH CEREBRAL PALSY

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ABSTRACT

Respiratory inductance plethysmography is being utilized to monitor the respiratory patterns in children with cerebral palsy. This non-invasive technique does not require maximal respiratory effort and is therefore a viable technique with the cerebral palsy population where traditional spirometric techniques are not useful.

INTRODUCTION

Cerebral palsy is defined as early acquired and non-progressive brain damage in which the specific sign is dysfunctionate motor control. One of the systems affected but not having received much attention is the respiratory system. Although the lungs are passive organs, the respiratory system depends upon good motor control for optimal performance. Also the respiratory system is intimately coordinated with swallowing as the pharynx serves an obligatory respiratory and alimentary role (swallowing). In cerebral palsy children who exhibit trunkal spasticity or generalized athetosis, the respiratory pattern is often uncoordinated. That is, thoracic and abdominal/diaphragmatic patterns are asynchronous and cause breathing to be arrhythmic and shallow(4). This often predisposes these children to either chronic or recurring chest infections. These same children often have difficulty swallowing (giving up the airway) and clearing secretions since they also have a reduced tussive effort (2).

A hiatus exists in the literature related to respiratory functions in cerebral palsy. This is a difficult population with which to conduct traditional pulmonary tests since many children are unable to hold their breath or blow maximally into a mouth-piece(1). Respiratory inductance plethysmography (RIP) is an indirect and noninvasive pulmonary monitoring technique which does not require pronounced respiratory effort. It is a viable method of monitoring breathing patterns in cerebral palsy children and facilitates prolonged respiratory monitoring. The RIP monitors changes in thoracic and abdominal cross-sectional area, and, subsequent to calibration, provides information concerning tidal volume based on these compartmental movements(3). Abdominal (ABD) excursions predominate in the supine position and rib cage (RC) movements predominate in standing. Since RIP does not utilize a mouth-piece or face-mask, eating may be introduced into the monitoring process so that the coordination of respiration and swallowing may be assessed.

The goals of this research study were to develop and refine monitoring and calibration techniques for respiratory inductance plethysmography with cerebral palsy children; to monitor and create a data base of respiratory patterns; and to evaluate respiratory data in relation to airway control during swallowing.

SUBJECTS

Three groups of children aged 5-12 years, participated in this study. Group I consisted of 3 non-neurologically impaired children, Group II consisted of 3 children with spastic cerebral palsy; and Group III consisted of 3 children with athetosis. Children with marked intellectual retardation or with an upper or lower respiratory tract infection or chronic chest ailment were excluded.

INSTRUMENTATION

The respiratory inductance plethysmograph (RESPITRACE[A]) consists of two transducers, parallel coils of Teflon insulated wire sewn in a zig-zag fashion onto two separate elasticized bands approximately 10 cm in height encircling the rib cage and abdomen(3). Respiratory movements elicit changes in the cross-sectional area of the rib cage and abdominal compartments and thus alter the self-inductances of each of the coils. This change in inductance is measured by placing each coil in an oscillator circuit, and thus changes in the volumes of the enclosed parts due to respiration are indicated by changes in the frequencies of the oscillators.

Three channels of respiratory information (RC, ABD and RC + ABD (Tidal Volume)) and three channels of electromyographs (EMG) (masseter, infrahyoid and diaphragm) were recorded simultaneously. EMG signals were monitored and preprocessed by an Autogen 1700 Myograph. The 100 to 200 Hz bandpass was selected for the infra-hyoid and masseter channels, with the signals being full-wave rectified and integrated over 50 ms. This facilitated peak parameter determination, such as timing (onset and fall-off) between different muscle peaks, peak widths (seconds), amplitudes and general trends. These parameters were correlated to respiratory patterns on a time sequence analysis. Diaphragmatic EMG was monitored by a set of bipolar electrodes located just inferior to the anterior aspect of the lower costal margin at approximately the level of the lowest point of the 9th rib. The EMG was filtered to pass signals from 100 Hz to 1000 Hz.

The signal analysis for masseter and the infrahyoids group of muscles was done using a Tektronix 4052 desktop computer. The Respitrace was calibrated by having the subject take eight breaths through a pneumotachograph via a facemask, first in a supine position and then in the subject's normal sitting (working) position. The rib cage coil was located over the sternum and under the axillae. The abdominal coil was placed midway between the lower margin of the ribs and the top of the hips, with the top of the coil just below the umbilicus.

METHOD

Three sessions each of approximately one hour's duration, including set-up time, were required. Prior to monitoring, a clinical assessment was conducted related to rate of respiration and any specific chest deformities. Each session followed this format: 1) quiet breathing until S is seen to be comfortable and at ease - not to exceed 5 minutes; 2) 3 minutes of monitored quiet breathing; 3) S asked to take as big a breath as possible and then relax; 4) repeat step (3) above; 5) One distinct sip of 5 ml. of water from a cup, followed by a swallow; 6) Repeat step (5) above; 7) Continuous drinking of 75 ml. of water; 8) Eating bite-sized piece (1 cm x 2 cm) of arrowroot cookie; 9) Repeat step (8) above; 10) Repeat step (7) above; and 11) 3 minutes of monitoring quiet breathing.

RESULTS

For the subjects of this study, a change in body position had very little effect on altering the abdominal (or diaphragmatic) component of tidal volume, while the major alteration although small in volume, occurred in the ribcage compartment. Normal subjects although exhibiting breathing patterns closer to the patterns described in the literature still presented data which did not vary significantly in different body positions. That is, no major shift from RC movements to ABD movements occured as the body position changed from standing to supine. This relative absence of shift in the respiratory pattern was contrary to results reported in the literature.

In the regular calibration routine, this effectively resulted in the derivation of the equation of a horizontal line in an RC/VOL versus ABD/VOL coordinate system, from which calibration values could not be derived. This difficulty in calibrating the Respitrace was solved by giving equal weighting to the RC and ABD signals for the recordings, while the spirometer or pneumotachographic signals were recorded simultaneously during time periods of 10-30 seconds for each session. Thus, relative changes although not in litres, could be observed in the coordination and change in contribution of each of the two compartments. Subsequently, by correlating the tidal volume (RC + ABD) signal from the Respitrace to the pneumotachographic signals, a scaling factor was calculated for the tidal volume (TV) signal, and hence, quantitative measurements of tidal volume could be made. The scaled Respitrace tidal volume signal remained accurate for the duration of each session. The resting breathing rate for the normal children averaged approximately 24 breaths per minute with a tidal volume of approximately 200 cc. In contrast, the resting breathing rate for the cerebral palsy children averaged 42 breaths minute with a tidal volume of 90 cc.

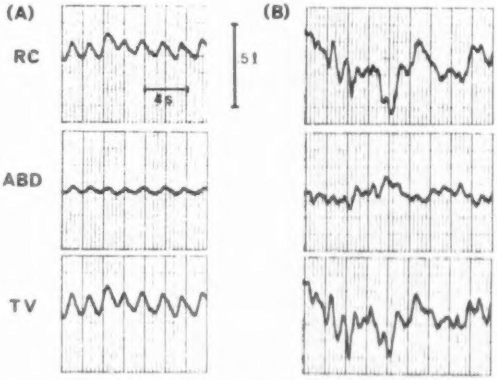

Figure 1: Baseline Respitrace tracings from a child with spastic cerebral palsy demonstrating the type of variation which can be observed during two periods of quiet breathing: (A) good rhythmic and phasic correlation; (B) poor phasing and correlation between Ribcage (RC) and Abdomen (ABD). (TV = Tidal Volume)

While performing the continuous task of drinking, the normal children were able to interspace an occasional breath as required. Cerebral palsy children were unable to do so and continued to swallow without breathing for as long as the stimulus to drink (the cup against the lower lip) was present. Figure 1 presents a sample of recordings from a child with cerebral palsy during quiet breathing and demonstrates the type of variation that one subject can exhibit.

In addition, this work confirmed that children with moderate to severe cerebral palsy are unable to hold their breath or to generate sufficient force to breath into a bell-type spirometer and lift the bell.

CONCLUSION

It appears that inductance plethysmography is a viable technique for monitoring respiratory patterns with cerebral palsy children. The objective monitoring of respiratory patterns will facilitate the evaluation of therapeutic interventions utilized to increase depth of breathing and correct asynchronous patterns. The effects of adaptive seating and chest restraints on respiratory function may now be assessed. In addition, objective, non-invasive information related to the coordination of respiration, deglutition and swallowing is now possible and may impact on feeding techniques. A large subject population needs to be studied but ultimately this work should be applicable to populations such as quadriplegic spinal cord injuries and acquired head injuries who also experience respiratory dysfunction.

A. RESPITRACE®, Ambulatory Monitoring Inc., 731 Saw Mill River Road, Ardsley, N.Y., 10502, U.S.A.

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ISOMETRIC MUSCLE STRENGTH AND ENDURANCE MEASUREMENTS USING AN APPLE COMPUTER

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ABSTRACT

An Apple computer was used to develop a system to quantitatively measure isometric muscle strength and endurance. The system is interfaced with a joint stabilizing chair or table that can use up to 8 load cells to measure the strength of 24 muscle groups. The delay time, rise time, and decay of the force response are calculated. The EMG is measured and parameters related to the timing of the EMG with respect to the force response are calculated. The system provides a cost effective solution for quantitative muscle strength measurements.

INTRODUCTION

In current medical practice the strength of a patient is determined by subjective measurements made by the examiner. The ratings are usually on a scale of 0 to 5. A grade of 5 is normal. A grade of 0 is no function and 1 is a trace, therefore, only numbers 2 to 5 are used to report changes in measurable strength. This does not allow for an accurate evaluation to be made or for quantitative determinations of changes that occur due to therapy or drugs. The purpose of our research is to develop a low cost, computer based, isometric strength and endurance testing system that applies to 24 muscle groups. By computerizing the measurements, both the temporal and amplitude factors can be categorized which allows for the possibility of determining muscle fiber type and fatigue characteristics. It also allows for a determination of the relationship between electrical and mechanical events.

SYSTEM CONFIGURATION

The system was designed around an Apple IIe computer. This computer was chosen because of its low cost and ability to accept an analog to digital converter (A/D). The system was designed using an Interactive Structures model AI-13 16 channel, 12 bit A/D converter. The sampling rate is controlled by a California Computer Systems model 7440 programmable timer. The sampling rate for each parameter is 1000 samples/second but only the average of each 10 samples is saved. An interface module was designed that provides 8 instrumentation amplifiers for the load cells and 2 EMG amplifiers, one of which has a full wave rectifier and averager (6 ms time constant). There is also a heart rate channel that accepts either ECG electrodes or an ear pulse sensor.

The system is designed to measure 24 different muscle groups. This requires a special chair and table with multiple load cells that will stabilize the subject in order that only the desired muscle group will be measured. When the desired muscle group is chosen on the menu the proper load cell associated with the measurement is activated. The stabilization chair and table will not be discussed further in this paper.

PARAMETERS MEASURED

The measurements are made for either maximal strength or endurance. For either measurement the patient's action is directed by a tone from the computer. Some of the measurements relate to the subject's response to the tone.

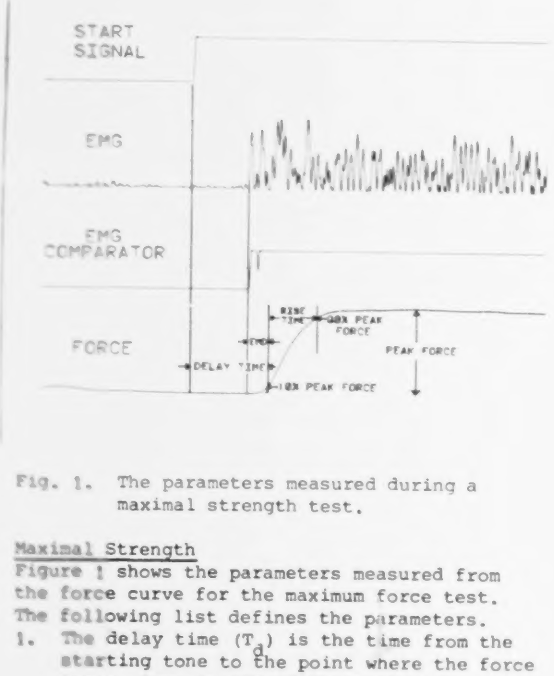

Fig. 1. The parameters measured during a maximal strength test.

Maximal Strength

Figure 1 shows the parameters measured from the force curve for the maximum force test. The following list defines the parameters.

1. The delay time (T_d) is the time from the starting tone to the point where the force curve rises to 10% of the peak force value.

2. The rise time (T_r) is the time the force rises from 10% of the peak force to 90% of the peak force value.
3. Electromechanical delay (EMD) is the time from the start of the EMG to the point where the force rises to 10% of the peak force value.
4. The peak force (F_p) is the maximum force occurring during the measurement period.
5. The end force (F_e) is the average force occurring during the last 1 second of the measurement period.
6. The average force (F_a) is the average force over the entire measurement period.
7. The linear slope of the force decay is measured from 800 ms after T_d to the end of the contraction.

Endurance

The endurance program makes measurements at a submaximal force level for continuous contractions or periodic work-rest cycles. The patient is shown a line on the screen which represents a percentage of his/her maximum strength. The patient is instructed to produce a contraction which moves a dot up to the line each time a tone is heard. The stopping conditions can be either a fall in the force to a given level or a chosen time duration. The force parameters measured are the following:

1. The average force over the entire run.
2. The duration of the run.
3. The slope of the decay of the average force as a function of time.
4. Data on individual contractions.
    a. Peak force (F_{pm}).
    b. Average force during the tone signal (F_{am}).
    c. Maximum force (F_{dm}) averaged over the length of time of the signal tone as a function of a delay of from 40 to 400 ms. This attempts to account for the delay in the starting and stopping of the contraction.
    d. The delay time, T_{dm}, at which F_{dm} occurs in part c.
    e. The electromechanical delay (EMD).
5. The EMG is sampled up to 10 times with .5 second periods during the measurement period and stored for later analysis.

SOFTWARE DESIGN

The software was designed to be user friendly and menu driven. It provides for entering patient descriptive data that is automatically attached to the quantitative data. The system requires 2 disk drives. One is used for the programs and the other for data. When the system disk is booted all the parameters and programs that are used are read into the extended 64K memory which looks to the operating system as another disk drive. This allows for very fast exchange of programs and storage of data. After the system has been loaded, the display shows the patient records that are saved by name, date, time, and the amount of space remaining on the disk in terms of the number of additional tests that could be saved. At this point the user can either print out existing data and/or delete records. The user is then asked for patient descriptive data such as name, age, weight, etc. The user next chooses disease descriptors from a major category selection of 21 types and then further describes the disease from a subcategory list that has a total of 158 entries. The selection of the muscle to be tested is done next, which also automatically selects the correct load cell for the measurement.

The next series of menus relate to the parameters of the test. For the maximum strength test the following parameters can be selected:

1. Maximum force range.
2. Duration of the test (2.5, 5, or 10 seconds).
3. Measurement of electromechanical delay.
4. The use of a preload on the muscle. If chosen, the computer will measure the changes from the preload value and record the value of the preload.

For the endurance test the following parameters can be selected:

1. Desired working strength level expressed as a percentage of the maximum strength.
2. Stopping conditions (time or force level).
3. Measurement of electromechanical delay.
4. Measurement of EMG data for frequency analysis.

After the user selects the desired option a graph is shown that displays time and amplitude axes. Upon pressing return a tone is sounded that directs the patient to contract the desired muscle.

DISCUSSION

The described system will allow clinicians to obtain quantitative profiles of the muscle strength of 24 groups in the body. Most equipment available focuses on the knee or only a few other muscle groups. Besides providing a very understandable parameter like maximum muscle strength, the system also provides a number of measurements related to the temporal response and also to the EMG. The clinical usefulness of the other parameters remain to be determined but offer potential to better quantify a patient's motor ability and give diagnostic information. From the initial design every effort was made to use low cost components in order to make the final product as economical as possible.

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ENGINEERING IN THE REHABILITATION OF DYSPHAGIA PATIENTS

11.10

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ABSTRACT

Rehabilitation of Dysphagia patients, in current rehabilitation practice, depends upon qualitative assessment of patient tissues often leading to trial and error. We have developed instrumentation to quantify the strength of the concerned tissues in terms of several biomechanical parameters which would significantly aid the rehabilitation process.

INTRODUCTION

Dysphagia is a disorder of the swallowing mechanism resulting from neurological impairment and presents a major problem in the comprehensive rehabilitation of the stroke patients and others with neurological disorders. Dysfunction of the swallowing process occurs as a result of lesions in certain cranial nerves, their nuclei, and fiber tracts or in the cortex. Due to the lack of quantitative measurement procedures, clinicians in the current practice base their judgment on the "feeling" of strength, of the concerned tissues, very often leading to "trial and error," and tedious course of recovery and rehabilitation. Quantitative measurement procedures are necessary for clinicians to better understand the recovery process and to prescribe therapy to achieve maximum rehabilitation and recovery rate. We have identified and developed technique and instrumentation to noninvasively quantify several biomechanical parameters that characterize the dysphagia patient.

METHODOLOGY

The process of swallowing occurs in three phases: (1) an oral or buccal-pharyngeal phase involving the muscles of the lips, the cheek, the tongue and the oropharynx, (2) a pharyngeal phase involving the pharynx and the larynx, and (3) an esophageal phase. Most of the dysphagia patients have disorder of the oral phase and in certain cases the pharyngeal phase. We have made the following biomechanical measurements to characterize the muscles involved in oral and pharyngeal phases of swallowing: (1) lip interface pressure (compressive stress), (2) lip interface pulling force (shear force), (3) lateral, forward, and upward thrust exerted by the tongue, (4) predental suction pressure exerted on a straw, (5) post dental suction (swallow) pressure at the base of the tongue during swallowing, and (6) acceleration of the throat tissues during swallowing with an accelerometer placed on the outside of the throat.

The lip interface pressure is measured with a semiconductor pressure transducer (Entra Devices, Inc.). The lip interface pulling (shear) force is measured via a plastic beam through an ultraminiature load cell (Sensotec, Inc.). The patient is asked to exert a pulling force on the flat end of the plastic beam. The signals from load cell and the interface transducers are amplified and recorded (Gould). The tongue thrust is measured using a small straingaged spoon-like plastic beam. The static strain on the beam is measured using a battery operated strain indicator (MicroMeasurements, Inc.). The suction pressure is measured with a hydraulic pressure transducer (COBE) connected to a catheter. The signals from the hydraulic pressure transducer are amplified and recorded (Gould, Inc.). Power to the amplifiers and the recorder is drawn through a patient isolator.

RESULTS

We have found significant differences in the above parameters measured from normal and dysphagia patients (Fig. 1-2 and Table 1). Also, in unilateral paralyzed individuals, we found significant differences between measurements from normal and paralyzed sides. With these measurements we are able to characterize the degree of recovery in the swallowing mechanism. The quantitative measurements of the strength of the concerned tissues allow the clinician to prescribe better rehabilitation procedures. Moreover, these measurements can be used for training the dysphagia patients. Our hypothesis is that feedback of the quantitative biomechanical parameters to the patients will aid in achieving quick recovery.

Table	Normal	Paralyzed
Lip Closure Press. (mmHg)	158.71	45.45
Lip Interface Shear Force (gm)	222.75	1.00
Predental Suction Press. (mmHg)	130.73	31.02
Tongue Thrust) (gm)	506.66	152.06

These results represent the first quantitative measurement of the muscle strength involved in the swallowing process in normal subjects and in dysphagia patients. The oral phase of swallowing involves lip closure followed by placing of the food at the uvula through a coordinated action by the muscles of the tongue. The lip closure pressure and lip interface shear force measure the strength of the lip and cheek muscles. Most of the dysphagia patients cannot completely close the lips resulting in food spilling. Dysphagia patients often have either a unilateral or a bilateral loss of tongue strength. These facts are clearly evident from the above results. The biomechanical factors devised above, when measured from the onset of injury, provide a quantitative picture of the recovery process.

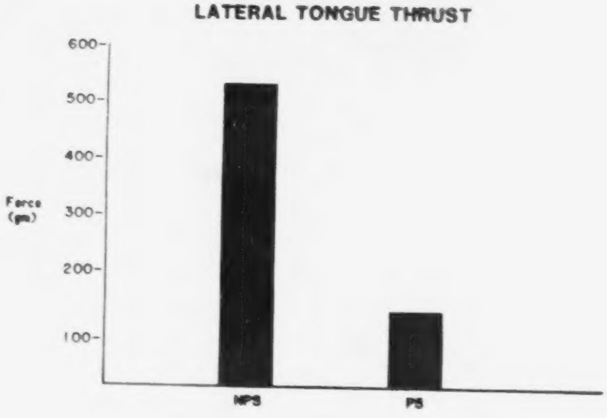

Fig. 1. The lateral tongue thrust exerted in normals (NPS) and Dysphagia patients (PS). These results represent the average of five repeated measurements from ten normal individuals and eight dysphagia patients.

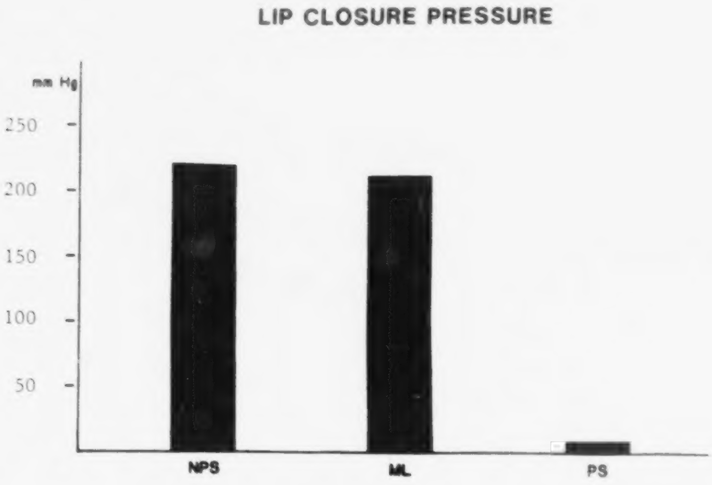

Fig. 2. The maximum lip closure pressure (interface compressive stress) observed on nonparalysed side (NPS), in the middle (ML) and on paralysed side (PS) of unilaterally paralysed dysphagia patient. These results represent the average of five repeated measurements from ten normal individuals and eight dysphagia patients.

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PRIORITIES IN RURAL REHABILITATION

12.1

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INTRODUCTION

Although the rehabilitation process is basically the same in all parts of the world, several factors serve to set apart the delivery and maintenance of the rehabilitation process in rural areas as compared to urban settings. Some of these are as follows:

- Disabilities unique to rural vocations, particularly farming (i.e. amputations due to farm equipment accidents).
- Unique accessibility problems
  - mounting a tractor
  - traversing soft uneven terrain in a wheelchair
  - architectural barriers in farm buildings
- Isolation factors which result because of
  - distance to social services
  - distance to primary and rehabilitation care facilities
  - lack of attendant care resources
- Reduced vocational and recreational opportunities
- Limited access to service i.e. down time for equipment repair etc.
- Expenses related to increase transportation and communication costs

Until recently, there has been little formal recognition of these factors. Within the past year, a movement has developed within the rehabilitation community to assess and document the impact of rural environments as they require unique rehabilitation services and technologies. One example of this movement is this very forum, RESNA's Special Interest Group on Rural Rehabilitation. Another example is the International Conference on Rural Rehabilitation Technologies (ICRRT) held at the University of North Dakota (UND) on October 23-25, 1984.

This paper summarizes the background, goals, and results of that conference.

BACKGROUND

The University of North Dakota is located in Grand Forks, North Dakota, in the heart of the agricultural Red River Valley. For the past twenty-seven years, the University's Medical Center Rehabilitation Hospital (MCRH) has served the Upper Midwest region extending from Montana to northeastern Minnesota, a region in which over 50% of the population lives on farms or in towns of less than 2500 population. Throughout the history of the MCRH, practitioners have been faced with the problems of distance between consumers and services, the lack of support systems, the unique problems created by a combination of the rural environment and the extreme climatic variations of the area, and the special needs of disabled farmers. Through the years, several joint research and development efforts have been undertaken with the UND Engineering Experiment Station to seek technical solutions to some of these problems.

The concept of a conference to focus on the specific, special needs of the rural disabled was first suggested in a UND proposal to NIHR in 1981 as part of a response to an RFP for a rehabilitation engineering center. While the proposal was not accepted, the idea of a conference was strongly endorsed. After reviewing the concept with several practitioners from around the country and receiving further encouragement, planning was begun in 1983 for an international gathering of practitioners, laypeople, and consumers relative to rehabilitation practice in rural areas.

METHODS AND APPROACH

The conference was organized along the themes of Consumer Characteristics, Aspects of Living, Service Delivery Models, and Technologies for Rehabilitation.

Consumer Characteristics, for special populations, focused on the demographic and diagnostic factors that describe certain populations living in rural areas. Studies which considered different age groups, types of disabilities, and environmental effects were sought.

Aspects of Living looked at independent living programs, recreational activities, and employment factors in rural areas.

Service Delivery Models centered on methods by which individual care providers or organizations, both public and private, attempt to respond to the needs of disabled persons in rural settings. Developments in mobile units, hospital and social service outreach programs, and shared service operations were invited.

Technologies section looked at equipment, appliances, or other devices and systems that are needed or have been developed for disabled people in rural places.

In addition, a separate workshop was planned for the disabled farmer. This workshop, conducted by Dr. Bill Field of Purdue University, was designed to present technical developments in making farm equipment more accessible to the physically disabled. In addition, a panel comprised of active farmers who have experienced some form of disability during their career were invited to share their experiences in coping with the new challenges of farming with a disability.

The conference presented an opportunity to document the interests of the rehabilitation community and laypeople with respect to their information needs. Registrants completed a conference evaluation form at the close of the conference. In addition, a post conference survey was sent to all registrants to aid in the development of future conferences. Together these evaluations provide a foundation for developing future programs and projects to meet the needs of the rural disabled.

RESULTS

The conference brought together more than 200 registrants and 40 papers in the four theme areas. Representatives of Sweden and India joined the predominantly Canadian and U.S. audience with twenty-four states represented. Contacts with rehabilitation professionals in Europe, Africa, the Middle East and other parts of the world were established in the course of the conference development. The strongest participation, as professional groups, came from vocational rehabilitation counselors and individuals involved with independent living centers around the United States.

A total of 62 registrants completed the conference evaluation form and 60 individuals (nearly 30% of the attendees) responded to the post conference survey. Based on these sampling instruments, ICRRT registrants indicated the following order of priority within each theme area for topics which need to be addressed in future programs or services for the rural disabled.

ASPECTS OF LIVING

* Employment opportunities/workplace modification
* Independent living programs
* Communication and transportation problems
* Recreational opportunities
* Activities of daily living

SERVICE DELIVERY MODELS

* Shared service programs between rehabilitation centers and small community hospitals
* Outreach programs from rehabilitation facilities
* Mobile unit service programs
* Satellite operations of comprehensive rehabilitation care facilities

CONSUMER CHARACTERISTICS/SPECIAL POPULATIONS

* Specific physical disabilities and their unique requirements in rural settings
* The multiple-handicapped
* The sensory impaired
* The disabled farmer
* The developmentally disabled
* Handicapped children in rural areas
* The elderly
* The chemically dependent

TECHNOLOGIES (Top ten interests)

* Occupational adaptations
* Transportation/mobility devices
* Daily living aids/adaptive devices
* Independent living aids
* Treatment services/delivery
* Farm site modifications
* Farm equipment modifications
* Computers
* Architectural barrier removal
* Prosthetics/orthotics

Among some of the other interests expressed where concerns regarding funding for rural rehabilitation programs and the dissemination of information of rehabilitation practices in rural areas. The need to further characterize rural environments and their effects on rehabilitation practices was also strongly indicated.

CONCLUSIONS

The first International Conference on Rural Rehabilitation Technologies confirmed the contention that rural environments present unique challenges to rehabilitation practices and the delivery of rehabilitation services. The participation at the conference and the strong follow-up interest indicates the need for some form of support group for both the professional and the layperson associated with rehabilitation services in rural areas. ICRRT II will be held at the University of North Dakota on October 22-24, 1985. Offers have already been made both from within the United States and abroad to host ICRRT conferences in 1986 and 1987 at other locations. The activities of the RESNA Special Interest Group in Rural Rehabilitation together with an ongoing series of ICRRT conferences will provide appropriate forums for networking the individuals concerned about the problems of the rural disabled.

REFERENCE:

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THE NATION'S FIRST MOBILE REHABILITATION ENGINEERING UNIT

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ABSTRACT

The W.W.R.C.'s Rehabilitation Engineering Services (RES) has provided both Center-based and community-based services for the past eight years. Since its inception, RES has dramatically broadened its horizons in a consistent effort to provide the most cost effective distribution of its services to the constituents within Virginia's Department of Rehabilitative Services (DRS). A capital investment which yields gainful employment for one of DRS's clients is a good investment, particularly since statistics prove that for each dollar Virginia's DRS invests in a client, the same client will return Ten as a taxpaying employee.

INTRODUCTION

The grass-roots involvement of RES staff in work station designs and enhancement has yielded hundreds of employment options that otherwise would not have existed. In a recent 8 month span alone, fifteen job site visits were referred to RES which resulted in 12 lasting employments due to custom-matched adaptive equipment manufactured and installed by RES staff.

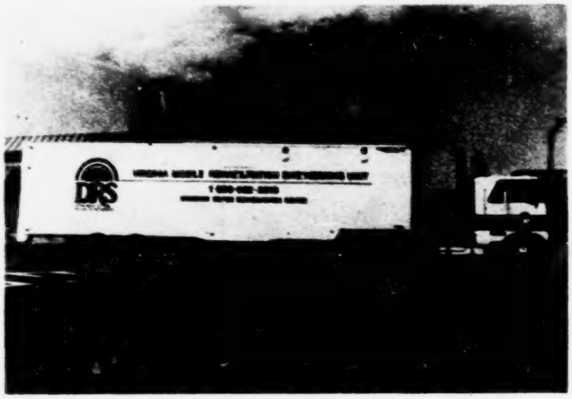
BACKGROUND

The geographical location of DRS's two RES facilities has necessitated WWRC's engineers serving an area greater than 30,000 sq. miles which extends from Tennessee to Washington, D.C.

With clinic and fabrication facilities located at WWRC, the actual cost of field services are unavoidably inflated by the number of trips to and from the jobsite as assistive devices are made, tested, and occasionally revised in our home-based laboratory.

Also requiring RES services are the over 500 clients residing at WWRC and participating in rehab. programs on a daily basis. This, combined with the ever-increasing necessity for service delivery in the field, demands that the limited RES staff time should be effectively used in "PROBLEM SOLVING" and not wasted traveling to and from a particular site to complete a task.

The only logical solution would be to take the shop along when staff must work away from home-base.

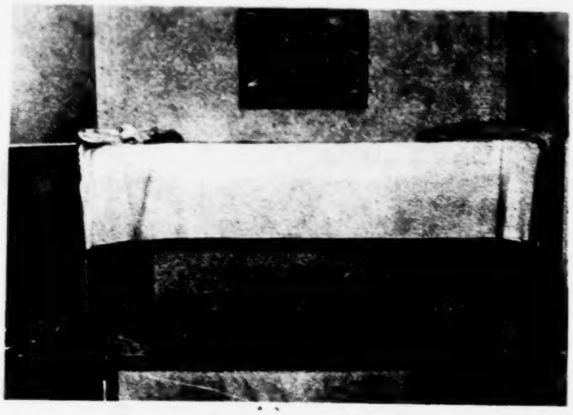


METHOD

An innovative grant proposal was submitted to DRS, and a $40,000 block grant was approved to purchase and modify a tractor-trailer into a mobile fabrication unit. Realizing that this would in no way cover the total cost, it was left to the RES staff to secure the balance of equipment, tools and materials through private and corporate donations. The actual conversion work was supervised and carried out by RES staff, with assistance from a contracted hourly employee and WWRC Maintenance staff.

The unit contains two 12-KW diesel power generators which supply the broad array of equipment housed within, that include heli-arc welding, lathe, mill, air compressor, bandsaws, drill press, glass bead finishing, metal shear, brake, roll former, wheelchair lift, lights and heat.

Also stocked on board are fabrication supplies and materials, hardware assortments, hand and power tools, etc. Two-man bunking quarters are found in the very front of the trailer which allow for no-cost overnight accommodations complete with refrigerator, range, sink, toilet, dining table, mobile phone service and T.V.



CURRENT UTILIZATION AND BENEFIT

At present, The Mobile Rehab. Engineering Unit is being allowed one week per month for field utilization. This is due partly to staff limitations, but actions are being taken which hopefully will alleviate this problem and thus allow appropriation of more staff involvement for the communities' needs.

The following is a list of paramount benefits of this type of service.

(1) To the Consumer:
- ° Allowing for greater rural access to the potentials of RES.
- ° Decreasing costs related to travel and incidentals.
- ° Decreasing the time to implement a solution.
- ° Preventing a more serious disabling condition from developing.

(2) To State Agencies & Funding Sources:
- ° Reducing costs by clustering visits.
- ° Increasing this agency's referral base thru non-DRS support.
- ° Fostering more interaction and growth between DRS and other health providers.
- ° Allowing placement couselors a better bargaining tool with prospective employers.

(3) To Business & Industry:
- ° Offering a support system and resource for affirmative action plans.
- ° Allowing for immediate "de-bugging" of equipment with interaction from both employer and client.
- ° Allowing for "on-Site" training of prospective new employees.

These benefits can be further substantiated by the following table:

Benefits of Mobile Unit

Parameter Measured	Traditional Service Approach	With Mobile Unit
Number of trips	3	1
Total distance traveled	759 miles	253 mi
Driving time for RE staff	13.8 hrs.	4.6 hrs.
Time between referral date and implementation date	3.5 months	3 wks.
Actual elapsed fabrication time	3 months	4 days.
Incidental charges incurred on trips	$119.00	$40.00

CONCLUSION

It remains to be documented just what the actual overall savings of such a unit would be, however it is estimated that the establishment of a stationary fabrication facility with the same equipment would be at least double what the mobile unit cost. That, coupled with the appealing ability to take Rehabilitation Engineering Technologies right into the rural areas (farms, small businesses, etc.) within Virginia, was the motivating force behind this project.

ACKNOWLEDGEMENTS

DRS
REHABILITATION ENGINEERING
EQUIPMENT DESIGNS & ADAPTATIONS FOR PERSONS WITH DISABILITIES

We gratefully acknowledge contributions by the following, without whom this project would have never materialized:

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ECONOMIC LOSSES ASSOCIATED WITH FARM-RELATED DISABILITIES

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ABSTRACT

The primary goal of this project was to develop a reliable method of computing and projecting the economic losses associated with farm accident-related permanent disabling injuries. It was decided that a computer based model would provide the most consistent and reliable method of computing and projecting the magnitude of losses associated with farm-related accidents. The IBM PC (Personal Computer) was selected upon which to develop the farm accident cost program because of its widespread use and availability. In addition, several other companies manufacture computers capable of running IBM PC compatible software.

INTRODUCTION

In 1982, the National Safety Council estimated that accidents cost the United States at least 88.4 billion dollars, of which 31.4 billion went to cover work-related accidents (Accident Facts 1983 Edition, 1983, pp. 4). According to Pfister (March 1983) "agricultural work accidents cost over 2.5 billion dollars per year, based upon current National Safety Council data". In other words, agricultural work accidents accounted for approximately 8 percent of the losses resulting from work-related accidents even though only 2.4 percent of the nation's population reside or work on a farm (U.S. Bureau of Census, 1982).

The costs associated with disabling injuries are responsible for a significant proportion of the cost of accidents. National farm accident data indicates that approximately 2 percent of the full-time farm operators and workers have suffered permanent disabling injuries due to farm-related accidents (Accident Facts 1977 Edition, 1977). A summary of farm accident data from 31 states, completed by the National Safety Council, found that 65.3 percent of all farm work-incurred injuries were considered severe injuries (Hanford, Burke, Fletcher, Hoskin and Miller, 1982, pp. 23). Approximately 1 percent of the farmers who experienced a non-fatal injury incurred a permanent disabling injury which prevented them from continuing to farm (National Safety Council, October 1981). The cost of these farm-related disabilities have never been investigated. Thus, it is the purpose of this paper to report on work being conducted by the authors to estimate the economic losses associated with permanent farm-related disabilities.

THE NEED FOR RESEARCH

A review of agricultural accident data indicates that very little has been done to determine the economic impact of farm-related injuries on the farm family or rural community. A few studies have briefly dealt with the cost of selected farm-related accidents and injuries. Stout and Darbee (1972) analyzed a few of the direct and indirect costs associated with 60 selected farm-related accidents and injuries, most of which were amputations. The National Safety Council summarized farm accident data from 31 states which included findings on the cost of farm-related accidents and injuries (Hanford, Burke, Fletcher, Hoskin and Miller, 1982). However, this data contained very little in-depth economic data on disabling accidents.

The major emphasis of previous farm accident studies has been to determine the frequency and types of farm-related accidents. Little effort has been devoted to identifying the economic consequence of farm-related accidents and injuries.

There are several important reasons for investigating the economic scope of farm-related accidents. These include:

1. The need to identify those types of farm-related accidents which are most costly to society and create the greatest economic threat to the farm family.
2. Current cost data does not provide a realistic picture of the losses incurred nor does it provide a means of comparing farm-related accident and injury costs with other industries or occupations.
3. The need to acquire concrete economic evidence to justify the expenditure of public, corporate, and private funds in loss prevention programs.
4. The need to develop fair and equitable guidelines for the allocation of economic liability in litigation resulting from farm-related accidents and injuries. A better understanding of the costs of farm accidents would assist in establishing more consistent and realistic award levels.

DEVELOPMENT OF A COMPUTER MODEL

It was decided that a computer based model would provide the most consistent and reliable method of computing and projecting the magnitude of the losses associated with farm-related accidents. The rationales for this decision were as follows:

1. A computer model allows for easy updating of the variables to account for inflation and future research findings.
2. A computer model will insure that a set procedure is followed.
3. After entering the various costs, a computer can quickly provide a visual summary of the costs incurred.
4. A computer program insures that the summaries obtained are consistent in format from one user to another allowing for comparative studies.
5. The computer can quickly cross reference data.
6. The computer would allow for the easy storage and retrieval of a large amount of data to allow for long term research to be conducted on farm accident costs.

The IBM Personal Computer (PC) was selected upon which to develop the farm accident cost program because of its widespread use and availability. In addition, several other companies manufacture computers capable of running IBM PC compatible software. Thus, the farm accident cost program would run on several computers in addition to the IBM PC. It was anticipated that the model developed would have application in any state to enable farm safety leaders and rehabilitation professionals to estimate farm accident losses.

COMPUTER PROGRAM COMPONENTS

Prior to writing the ACCICOST (name given to program being developed) computer program, the components of the program had to be identified. Based upon the review of literature and discussions with farm accident victims seven major components were discovered to be essential elements of the computerized farm accident cost program. These were: CPI (Consumer Price Index) Data; Personal Data; Body Part Injured; Types of Injuries; Injury Severity Level; Types of Accident/Injury-Related Costs; and Default Values. The following is a brief description of each major component and the role it plays in the ACCICOST computer model.

1. CPI Data

CPI values are necessary in order that the cost-related default values stored in the computer's memory may be updated to account for inflation and other cost changes. Determination of the CPI categories used was based upon the type of default costs used throughout the program. CPI categories used in the program include: hospital service; physician fees; auto repair and maintenance; housing maintenance and care; all items; and the state and local government workers employment cost index.

If the user decides to enter new CPI values, the computer will ask for the current date (month, day and year) following the input of the new CPI values. The date is used to inform ACCICOST users when the CPI values where last updated, thus, allowing the user to determine if the CPI values, currently in ACCICOST, should be updated.

2. Personal Data

Personal data requested includes: names of the injured person; date when the accident occurred; victim's age at time of accident; and victim's current age. Of the four personal data items requested, the only one of great importance is the victim's age at time of accident. The victim's age at time of accident is necessary in order that the computer may calculate the annual follow-up medical care and the productivity losses incurred.

3. Body Part Injured

The National Safety Council's farm accident reporting forms and reporting criteria from the American Association for Automotive Medicine were used to develop a list of 17 major body parts that can potentially be injured in a farm accident. The computer model will prompt the user to select the body part that was injured.

4. Types of Injuries

Form 2 of the National Safety Council's farm accident survey contains a section entitled, "types of injury". Injuries listed under this section include: amputation; asphyxiation; bruise; cracked, fractured or broken bones; cut or laceration; eye injury; mangled; pinched; puncture; and sprain. The American Association for Automotive Medicine also has compiled a list of injuries encountered due to accidents. Analysis of the two lists resulted in the selection of several injury types for each of the 17 body parts.

5. Injury Severity Level

In 1980, the United States Department of Transportation (DOT) published a study that analyzed the economic losses associated with motor vehicle accidents. The study entitled, "The Economic Cost to Society of Motor Vehicle Accidents" obtained most of its cost data from a 1975 study conducted by Hartunian, Smart, and Thompson (1981). As Hartunian, Smart, and Thompson did, the DOT study placed cost values on motor vehicle accident-related injuries by rating the injury's severity using the American Association for Automotive Medicine's, "The Abbreviated Injury Scale, 1980 Revision". The Abbreviated Injury Scale (AIS) classifies injury severity level on a numeric scale of 1 to 6. The severity code for the AIS code, as mentioned in the AIS handbook, is: 1 = minor; 2 = moderate; 3 = serious, 4 = severe; 5 = critical; and 6 = maximum injury virtually unsurvivable. Hartunian, Smart and Thompson (1981) expanded AIS category 5 into five subcategories: non-spinal cord injury patient; incomplete paraplegic; complete paraplegic; incomplete quadraplegic; and complete quadraplegic.

6. Types of Accident/Injury-Related Costs

The heterogeneity that is encountered from one farm accident to another makes it difficult to develop a comprehensive list of cost types. Based upon industrial accident studies such as Matthysen (1973), agricultural accident studies such as Robbins (1976), and visits with farm accident victims, the following list of cost types was developed: police assistance; fire department/rescue squad; ambulance service; hospital charges; medical expenses following leave from hospital; rehabilitation expenses; parking and mileage charges; loss of earnings; property damage; replacement labor; lost productivity; home and farm modifications; and legal expenses and or income.

7. Default Values

Default values enable the program user to project the cost if the actual cost was not readily available. The lack of certain types of farm accident-related cost date prevented the input of default values for each of the cost categories.

SUMMARY

ACCICOST will provide the user with an abbreviated and/or expanded summary. The abbreviated summary provides only a summarization of the major costs while the expanded summary contains a copy of the values inputted to the computer.

DEVELOPMENT OF ACCIDENT COST PROFILES

To obtain actual data on the types and magnitude of losses encountered by people who have suffered farm accident-related injuries on-farm visits were conducted. The primary objectives of the on-site visits were:

1. To determine the various types of costs encountered as the result of farm accident-related injuries.
2. To determine the magnitude of these various costs.

County extension agents and high school vocational agriculture instructors throughout the State of Indiana were contacted and asked if they could identify individuals in their community who had experienced a farm accident-related injury. In addition, clippings received from a newspaper clipping service were monitored in order to identify farm accident-related victims.

The data gathered from these visits were used to test both the input and output operations of the ACCICOST program.

POTENTIAL BENEFITS OF THE STUDY

In addition to fulfilling the needs addressed earlier, the authors see the following as potential benefits that are realized by completion of the study.

1. A comprehensive lists of costs associated with farm-related accidents and injuries would be identified.
2. A method of projecting farm-related accident and injury costs would enable estimates to be made on the total impact of farm-related accidents.
3. Government agencies would be able to identify the cost factors and the magnitude of the costs involved with accidents in order to better provide the financial relief required by individual victims.
4. Provide evidence to justify expenditures on loss prevention activities.
5. Help physically disabled farmers make decisions as to whether they should continue to farm following their rehabilitation.

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13.1

The Potential of Pitman Shorthand Transcription as an Aid for the Deaf

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Possum Controls

Summary

Machine Shorthand systems have been successfully used to provide simultaneous transcription of speech for the deaf in a number of environments. These, however, require a highly trained machine shorthand operator. In order to increase the potential of such systems, a study was conducted into the feasibilty of using handwritten shorthand as an input means. A prototype system was designed and constructed. The problems of translation of handwritten shorthand, however are very much greater than for machine shorthand, and the performance of the prototype system was not adequate for a working environment. Substantially more research work would be required to bring this up to a satisfactory standard. The performance of such a handwritten shorthand system however is unlikely ever to be competitive with a machine shorthand system as an aid for the deaf, but may possess advantages for a facsimile-based document editing system.

Introduction

A number of systems have been designed to transcribe the output from a shorthand machine into readable script. A major reason for this has been to improve the efficiency with which court transcripts can be produced. In addition systems of this type have been used as aids for the deaf. In such a system, the hearing person talks to a machine shorthand operator, who transcribes the words onto a shorthand keyboard machine. A computer translates this into readable script which is then displayed on a television screen for the deaf person to read. Such systems have been designed both for Palantype Machine Shorthand which is indigenous to the U.K., [1] and the American Stenograph system [2], and have been used in working environments on both sides of the Atlantic.

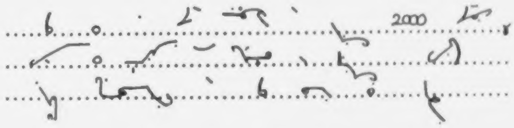
The most well known user in the U.K.is the deaf Member of Parliament Jack Ashley [3], and the Palantype based system has also been used by other deaf people at committee meetings, lectures, and as a prototype telephone service for the deaf. In America Stenograph systems are being used in educational institutions for deaf pupils, in law courts and as a television subtitling service for live programmes such as the news.

The above systems are working well and are commercially available. They provide an invaluable service to a section of the hearing impaired community, as they enable them to read a truly verbatim version of what is being said as it is being said. The major disadvantage of these systems, however, is that they do require a trained machine shorthand operator. In order to attempt to increase the pool from which operators of speech transcription systems could be drawn, we investigated the possibility of further developing our shorthand transcription system so that it could be used with handwritten shorthand [4].

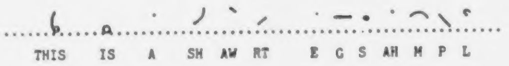
Handwritten Shorthand

There are a number of handwritten or "pen" shorthand systems, but the most popular one which can be used at high speed is Pitman's New Era, and we decided to concentrate on this system. An example of this shorthand is shown in Figure 1. It has a basic alphabet of 40 symbols representing speech sounds. These are simple lines, curves and circles which can be made with two pen pressures. Each word is written by a single geometric stroke representing the consonants within the word. Extra marks are then added to this base outline to represent vowel sounds. Common words can be represented by special "shortform" characters.

FIGURE 1 - an example of Pitman 2000 shorthand, illustrating the composition of a number of outlines.



This is a short example of Pitman 2000 shorthand. Work is currently in progress to determine whether computer transcription of this script is feasible.



Unfortunately pen shorthand is much less suited to automatic transcription than machine shorthand. A shorthand machine can be interfaced to a computer system simply by adding electrical switches to its keys - whereas a much more complicated system is required for pen shorthand. In addition to monitoring the movements of the pen, and inputting into the computer, the pen shorthand outline has to be recognised by the machine. This latter task is many orders of magnitude greater than detecting the closure of electrical contacts!

Unlike machine shorthand the phonetic data from pen shorthand includes word boundary data and thus the conversion of this data into an orthographic version is easier. A complication arises, however, because when taking down at speed many pen shorthand writers will leave out parts of the outline.

System Requirements

A pen shorthand system thus requires the following components:

1) A method of monitoring the pen movement and inputing this data into a computer system,
2) recognition of the outlines,
3) conversion off the phonetic outlines into an orthographic equivalent.

The development of an effective system is a very major undertaking but, rather than investigate one of these areas in detail and isolation, the authors decided to conduct a feasibility study which included the pilot development of a complete system. This development was thought unlikely to produce a system which would have an adequate performance but it would highlight the critical aspects of the problem and give an indication of the extent to which it would be possible to produce a working system in the future.

Data input and character recognition

The input of the original shorthand data did not present any insurmountable problems. A standard commercially available data tablet was found to have adequate resolution for pen shorthand shapes. It was not possible, however, to purchase a tablet which had sufficient resolution to detect the difference between thick and thin lines. We therefore added to the tablet an instrumented pen which detected whether heavy or light pressure had been used by the stenographer. The tablet and instrumented pen were interfaced to a Z80 based Cromemco computer system via serial lines.

Shorthand outlines thus inputed into the computer were represented as strings of digits corresponding to points on the data tablet. It was then necessary to recognise these strings as the phonetic characters which the writer had intended. This is a traditional pattern recognition task similar to handwritten character recognition.

This task was divided into a number of sub-tasks.

1. smoothing the data
2. segmenting the complex outline into its component parts
3. classifying these parts, and
4. recognising the shape and noting the relative position of vowel markers.
5. In addition to the above, a number of highly abbreviated shortforms are used for certain words and needed to be recognised separately.

Special pattern recognition algorithms had to be developed which were appropriate for pen shorthand symbols and programs were written to implement these algorithms. The data from the data tablet was thus processed by these algorithms to produce a string of characters.

Phoneme to orthography

The phoneme string now had to be transcribed into an orthographic representation. At this stage the data was in a similar form to that produced directly by a keyboard shorthand machine, except that it was divided into words rather than the syllablic structure inherent in machine shorthand.

Machine shorthand transcription systems use large dictionaries and, in some cases, spelling rules to perform this conversion. The dictionaries are an essential tool in reconstructing word boundaries. With pen shorthand, word boundaries do not have to be reconstructed, and also more sophisticated spelling rules which depend on position in the word can be developed.

However in a real environment, pen shorthand writers will often not transcribe vowels. This improves the speed and still retains a script that it is possible to decipher. This is a similar convention to that used in some speedwriting techniques where vowels are omitted and the words are disambiguated by a human being using syntactic and semantic information. NVRTHLSS IT IS NOT TRVL TO RD THS TPE F SCPT.

We thus developed spelling algorithms for Pitman shorthand and also a scheme for inserting vowel markers. There is not enough information in the phonetic data to reconstruct vowels which are omitted but it is possible to use statistical data concerning letter frequency distribution to calculate where vowels may have been omitted. In those cases we inserted a vowel marker to improve the readability of the text.

Prototype system

A complete system was produced in which shorthand data could be written on the tablet and an orthographic version produced. This proved that the overall concept could be realised. The performance of this system however was very poor. An analysis of the results of tests showed that the main reasons for this lay in problems of the initial recognition of the pen shorthand symbols. Although the primitive features of handwritten shorthand are very simple (lines, circles, hooks, etc,) even these simple strokes are not performed with sufficient regularity by shorthand writers to make recognition a straightforward task. More importantly it proved to be extremely difficult to accurately segment a complex outline into its primitive features.

Subsequent processing such as a phonetic to orthographic transcription and vowel insertion could be done to an adequate standard.

Operator Performance

We did not have the resources to perform large scale tests on the system and it is possible that substantial practice may well improve operators' performance, by teaching them to make their outlines less variable. There is a danger, however, that the need to write very precise outlines will cause an unacceptable reduction in the speed with which the shorthand can be written.

Conclusions

We have expanded our work on speech transcription systems for the deaf to include pen shorthand as an input medium. The results of this work however do not augur well for the production of a viable system which will compete with machine shorthand as a verbatim transcription aid for the deaf. Substantial research and development effort needs to be expended on the pattern recognition aspects of the system, particularly the segmentation algorithms. Although the systems may become usable, they are unlikely to produce an output which is of comparable quality to that from machine shorthand transcription and the speed with which operators can use it is likely to be less than that of machine shorthand. Nevertheless pen shorthand transcription offers a number of advantages for other applications such as facsimile-based document editing where a pen chould be used for all stages of the work.

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PAL: AN EFFORT EFFICIENT PORTABLE COMMUNICATION AID AND KEYBOARD EMULATOR

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ABSTRACT

A communication aid for the physically handicapped offers the user complete word predictions from a dictionary which is adaptive (modified through use). Implemented on the Epson PX8, the aid is truly portable, and when used as a keyboard emulator up to 60% saving in character selections is possible.

INTRODUCTION

One of the most basic but essential human needs is that of communication. For most people it is a skill mastered early in life and practised almost effortlessly thereafter, both in the form of the spoken, and written word. However, for those with some form of motor handicap, communication is a major problem. A number of microcomputer based aids have been constructed which provide an alternative means of input, and hence enable text to be created by those whose disability renders the traditional keyboard awkward (1,2,3,4). Due to the users handicap communication rate is usually constrained by factors such as motor response and physical fatigue and hence these aids are generally very slow to operate and do not exploit all the facilities available in a microcomputer.

Written English is a very redundant coding method from an informational view point, and it is therefore possible to enter a given piece of text by far fewer character selections than there are characters. A number of researchers have combined one of the alternative input devices with a form of 'text acceleration' which exploits this, such as: N-gram techniques (5), Coding (6), and Word prediction techniques (7,8,9,10). Some of the aids that have been produced use a combination of these methods (8,9,10). A predictive communication aid has been developed at Dundee university which offers complete word predictions based on the current prefix, and by so doing substantially reduces the keying effort needed to communicate (11,12).

THE PREDICTIVE ADAPTIVE LEXICON

The Dundee Predictive Adaptive Lexicon (PAL) was designed to be run on a portable microcomputer with a small visual display screen so that it could act as a keyboard emulator to a main microcomputer running word-processing or other packages. PAL could be controlled from the keyboard of the portable microcomputer or by a special interface such as a sip-puff switch with scanned matrix.

Prior to the user selecting a character, or after a word termination (space etc.) PAL offers the user a list of the most common words that it has in its internal dictionary. The user may select one of these by pressing a function key (if using the keyboard) corresponding to the position in the list of the word he wants. In this case the word chosen, along with a trailing space character, is forwarded to the target computer. If he is trying to type a word that is not in the list he must select the first letter. PAL then responds with a list of the most common words beginning with that letter. If the user sees the word he wants he may select it using a function key, otherwise he types the next letter. PAL responds with a new menu of the most common words begining with this two letter prefix. The process is repeated until the user selects a prediction or completes his word and terminates it with a space.

A substantial advantage offered by PAL is that the dictionary and statistical word usage data (frequency, recency) is acquired by the system as the operator uses the predictor, hence if the user types a word which is not yet 'known' it will be captured, entered into the dictionary, and will thereafter be offered as a prediction. The dictionary therefore adapts to the vocabulary of the user while in use. In order to give a more rapid change in predictions for a change in subject than would be possible through the evolution of a dictionary, the user may keep several context oriented dictionaries, and store them on disc when not in use. Each dictionary is constrained to a maximum size of the order of 1000 words (approximately 75% of common vocabulary (13)) in order that in use it may all reside in system memory. This restriction keeps machine search time at an acceptable level.

IMPLEMENTATION OF PAL

The Epson PX8 portable microcomputer was chosen as the basis of a prototype for PAL. The PX8 is battery powered with a CMOS Z80 as its CPU and 64 Kbytes of CMOS RAM. It has an 8 line by 80 column LCD display and an integral keyboard. The PX8 comes fitted with the standard CP/M operating system, which is housed in ROM. As an additional 'bolt on' unit there is a 120 K byte RAM disc which is treated as a fast access disc drive by CP/M programs. The PX8 also offers a microcassette tape drive for 'hard' storage and a fully configurable RS232 serial I/O port. Access is possible to the full Z80 bus through an edge connector, so any other form of I/O could be added if desired. The serial port on the PX8 may be used so that PAL can take its input from a sip-puff scanning matrix or from a Concept keyboard (an A3-sized membrane keypad with 128 separate zones with which keyboard overlays may be used) (14).

A disadvantage of the current version of the PX8 is that the legibility of the display panel is disappointing, particularly as it has a small viewing angle. Nevertheless it is adequate and one would expect subsequent versions of this and similar portable microcomputers to have improved visual display panels.

The PX8 version of PAL offers 5 predictions at any one time, and these are offered as a vertical list in frequency order with the most common word at the top. This appears to be the optimum layout for searching. A horizontal list was experimented with, to assist spatial mapping between each word and the corresponding function keys (which are layed out horizontally in front of the LCD display). No search time figures were measured, but the horizontal list was subjectively reported as being more tiring and more difficult to search.

The keyboard interface of PAL is configured to make the PX8 suitable for single finger operation by the use of 'soft' shift and control keys. With one finger it is possible to generate the full complement of control codes, which is essential if the aid is to be used with programs such as Wordstar running on the target system.

RESULTS FROM TRIALS

It has not yet been possible to carry out extensive long term trials with users, but those disabled people who have tried the system over short term periods have testified to the benefits that they experience in using PAL. We have however been able to calculate the reduction in keystrokes which would be produced in a realistic environment by PAL. We have developed a 'synthetic user' test in which pieces of text can be passed through the system and the number of keystrokes which would have been required to produce this text calculated. The main test piece ('Albans') is a transcript of a lecture on T.V. sub-titling for the deaf, this has the advantage of being of reasonable length (greater than one hour) and having the attributes of both ordinary spoken speech and a technical article. Other test pieces include various related newspaper articles and a large technical manual.

A simple algorithm was used in the synthetic user such that if the singular of a word is seen in a list, when in fact the plural is required, the singular form is ignored and the plural is typed in its entirety. A real user would probably select the word and after removing the trailing space will add the pluralisation. Hence a pessimistic figure of key saving is generated by the synthetic user.

Each test started with an empty dictionary. With ten predictions being offered, the newspaper articles generally yielded a 30% key saving (i.e. 7 out of 10 characters need be entered). The technical manual yielded a key saving of 46% and 'Albans' a key saving of 43%. It was found that the key saving had usually reached 25% within the first 250 words of the text.

When the number of predictions offered was reduced to 5, it was found that for 'Albans' the key saving dropped to 40%, and for three predictions the key saving fell to 38%. Reducing the number of predictions offered does not radically affect the saving because word frequency drops exponentially with rank order (15,16) hence the word required will generally be near the top of the prediction list.

If a text is processed with PAL using a dictionary already created from that text, a maximum key saving will be possible as all the words are already in the dictionary. This would roughly correspond to the user switching to a particular 'context' dictionary. This gave key savings lying in the range 58% to 62% for all the texts tested. As the average word length for the texts is just below 6 characters (5.53 for albans) this equates to 2 to 2.5 keypushes per word, one keypush for the first character, then one keypush to select a correctly offered prediction.

In order to try and optimise the key saving a model was proposed whereby word pairs (simple phrases) are also offered as predictions. Several texts were analysed and the word pairs extracted, the additional key saving that would occur was then calculated. The results were not encouraging with at best a 3% additional gain, this was due to two reasons. Firstly, in the majority of pairs one of the words was short, 2 or 3 letters, and of high frequency, hence we only save 1 or 2 key pushes through selecting it in a phrase. Secondly because the dictionary now contains all the word pairs and is therefore very much larger, it will take more key pushes to access some lower frequency words. On the basis of this it was decided to suspend the development of a phrase capture system.

CONCLUSIONS

A portable microcomputer based aid for the physically handicapped, which gives the disabled person access to most standard computer systems, has been developed. As well as giving a potential reduction in keying effort of over 40% it will accept input from any interface device which is matched to the users abilities. However although the system has been used in an academic and limited clinical setting, another task, as difficult as the development work now confronts us: that of arranging for it to be packaged and retailed at a price the disabled can afford so that it can be used by the people it was designed for.

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13.3

Introducing Microcomputers into Occupational Therapy - UK Experience

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Summary

Microcomputers were introduced into Occupational Therapy Departments in the U.K. in 1982. It has become clear that they have substantial potential, but there is a lack of both useful software and of guidelines for the use of this technology in therapy. A research programme has been instituted to tackle these deficiencies. In particular computer games technology was found to have potential, but in their standard form the games were inappropriate. We have thus started a programme of modifying games software, and producing guidelines for such modification, and for the maximization of the potential of Microcomputers in Therapy.

Introduction

In the United Kingdom, 1983 was designated as "Information Technology" year. As part of this initiative, the Department of Trade and Industry (a body funded by central government) produced a number of initiatives designed to encourage the use of information technology products - in particular microcomputers.

Following representations from the College of Occupational Therapists, it was agreed that this initiative should include Occupational Therapy. A number of steps were taken to achieve this; the major ones being:

a) Gifting microcomputers to thirty Occupational Therapy Departments in the United Kingdom,
b) Setting up an introductory training course, and
c) Arranging for servicing facilities for the microcomputers.

This initiative was designed to last for two years; it being hoped that it would be possible to assess the potential of microcomputers in Occupational Therapy during this period. The Microcomputers were delivered in 1983.

It was decided to standardise on one computer for this initiative; the BBC model B (marketed by Acorn Computers) was chosen, which was equipped with a colour monitor, disc drive and a touch sensitive keyboard (marketed by Concept Keyboards Ltd.). The initiative concentrated on the provision of hardware and no special arrangements were made to develop or distribute software.

A three day training course was mounted at the National Physical Laboratory (London), and two or three therapists from each Department attended these courses. N.P.L. were also made responsible for the servicing of the microcomputers.

Current Status

As the initiative drew to an end it was clear that it had produced a range of impacts in the Occupational Therapy Units. At one extreme the microcomputers have had little impact on the operation of the Department. At the other extreme, in a number of Departments, the staff have become very enthusiastic indeed, and the gift has been followed by the purchase of many more computers using charitable funds to which the Department had access, or which were solicited with a view to buying more computers (The increasing popularity of home computers during this period meant that this was a popular fund raising activity). This latter response was usually the result of there being a particularly enthusiastic therapist on the staff, often with access to someone - a friend or relative - having computer experience.

In addition to local initiatives of this nature, the College of Occupational Therapists set up a special interest group, and also began to publish a news sheet entitled "O.T. Micronews". This consisted mainly of short articles on therapists' experience with microcomputers, and also provided an information service on hardware and software availablity.

Survey

A survey has recently been commissioned by the Department of Health and Social Security (of England) to examine the uses to which these microcomputers have been put. In the D.H.S.S. survey it is suggested that microcomputers can potentially be used in a wide range of therapeutic regimes for:

Cognitive processes (concentration, memory, decision making, etc.),
Perceptual skills (shape discrimination, spatial awareness etc.),
Communication,
Motor skills (including hand-eye coordination),
Education,
Work assessment,
Stimulation of interest, diversion, and leisure, and
Social/life skills (including behaviour modification).

Examples of therapists using microcomputers in the above activities have been found in a pilot of the above survey, but many of the respondents comment on the lack of appropriate software.

Software availability

It is clear that a great deal of progress has been made, and that this technology has a place to play in therapy. A major problem, however, is the lack of appropriate software. Although there is a vast amount of general purpose software and some educational software for the BBC microcomputer, this is rarely suitable for therapeutic uses.

This lack of both software and infrastructure for software development posed a problem for which solutions were sought at a local level. Thus some therapists began to write their own software, or, more usually, solicited the help of relations, friends, "friends of the hospital", etc. A number of very worthwhile therapeutic regimes arose from this type of activity, but, as would be expected, it was not a co-ordinated activity, thus duplication occurred, and the quality of the software produced was variable.

Dundee District (Scotland)

Two microcomputers were delivered to Dundee District as part of the D.T.I. initiative. The District Occupational therapist approached the University of Dundee Microcomputer Centre for assistance, as this Centre was already active in the use of microcomputers as aids for the disabled.

The Microcomputer Centre thus organised half-day introductory courses for all the therapists in Dundee district (and also the surrounding Tayside Area). These courses were only meant to introduce therapists to microcomputers, and - more importantly - to try and allay their fears of these devices. One of the major problems experienced by new users is the fear of breaking the machine, and it was felt important to re-assure therapists that they were, in fact, very unlikely to damage the computers themselves, even though the software may well malfunction (The fact that young children do not have such inhibitions about breaking things goes a long way to explaining the popularity of microcomputers with this age group!).

On the basis of discussions at these meetings it was decided to mount a research programme with a view to exploiting the potential of and partially solving the problem of the provision of software. This research programme was to be located in the Microcomputer Centre, at Dundee University, and the work would be done by an interdisciplinary team of a therapist and microcomputer engineers. This research was funded by the Scottish Home and Health Department (a goverment agency, which is the Scottish equivalent of the English Department of Health and Social Security) for a period of three years commencing October 1984.

Research Programme

The S.H.H.D. supported research programme has four major aims:

1) To investigate the appropriate uses of microcomputers in Occupational Therapy,
2) To lay down guidelines for the use of this technology in therapy,
3) To develop examples of appropriate software, and
4) To lay down principles and guidelines for software developers in this area.

In our initial investigations, and following discussions with other Occupational Therapy Departments it was clear that the software and hardware requirements for useful therapy are much greater than commercially available systems can offer - even at relatively simple levels. Commercially available software, particularly the very popular computer games, however, provided a number of ideas which were certainly motivating and potentially therapeutically useful. For example "action" games were promising for improving dexterity, hand-eye co-ordination, etc.

Unfortunately the vast majority of commercially available games are not immediately useful, and in many cases would be positively harmful. Typical of their shortcomings are those very features which make them appealing to dexterous, mentally alert, able-bodied persons - they are fast, competitive and difficult; this, coupled with the extreme emphasis on success, a hall-mark of many games, is counter-productive for most therapeutic applications. In addition games rarely score the appropriate response for client-assessment, and often require permanent staff presence.

Writing good software, however, is a very time consuming process. We have thus concentrated initially on chosing games which can be modified so that they can be used in a therapeutic environment, and on what principles need to be adopted in such modifications. We have produced a suite of five programs based on computer games software which have been modified specifically for therapy with head-injury clients. The suite consists of:

Wordcross-therapy: searching for words in a grid (useful for co-ordination, spatial awareness and spelling),
Watchperson-therapy: a route planning game,
Colour-therapy: a colour matching task, including reaction time measurement,
Numeric Invaders-therapy: a space invaders look-alike,for hand eye co-ordination and numeric skills,
Reds and Greens-therapy: similar to the "Connect 4" game, to encourage spatial awareness, and forward planning.

The major modifications which were introduced were:

A) The instructions were simplified and extra "help" pages provided.

B) The games were modified so they required very much lower skill levels. This included slowing the games down, reducing the size of the problem presented, etc.

C) An "alter" facility was provided so that the skill level could be changed over a very wide range under the control of the therapist. This was a vital modification as the therapist could then choose the most appropriate level and also change this as the client improved (or got worse in the case of clients with degenerative diseases).

One particularly important part of the "alter" facility was that it was designed so that the skill level could be changed at any time without loosing the current status of the game. The client could thus be returned to the point in the game where he broke off. This is particularly important at the beginning of therapy where the skill level has to be set by trial and error, as it means that the client does not keep having to return to the beginning of his game with the attendant frustration this causes.

D) A further feature which has been added to this suite of games is a "return-prompt". Clients with short term memory problems were unable to play some games simply because they were unable to remember the traditional need to press enter/return. It is not always possible to remove this requirement, and even notices stuck on the screen did not help one of the clients.

This problem was solved by arranging for a flashing reminder to appear on the screen at the approporiate time. Not only did this mean that the client was able to play the game, but also it was found that the flashing reminder could be withdrawn after some time. Our games were thus provided with a "reminder" facility which could be turned off and also which could have a variable delay inserted between "press-return" being required and the reminder appearing.

E) To cater for clients with poor manual skills our games can also be operated by a touch sensitive keyboard and programmed to have large touch sensitive areas.

F) In addition, monitoring facilities are being added to these, and other games we are developing, to measure the time to respond, number of incorrect tries etc. This means that a profile of the clients' performance can be obtained, for assessment of both client and therapy.

Although it has not yet been possible to assess fully this software, pilot trials with clients have been very promising and augur well for the future.

Conclusions

Microcomputers can offer substantial benefits in therapy if the software is good and that the technology is being used in appropriate circumstances. In particular computer games technology has shown promise, but generally needs significant adaption before it is suitable for therapeutic uses.

On the basis of our work we are developing guidelines for modifications to software and for the design of novel software for therapeutic uses. Initial response to our work, particularly with head-injury clients, has proved very encouraging.

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ARTIFICIAL ELECTROTACTILE COMMUNICATION: DECOUPLING INTENSITY VARIATIONS FROM FREQUENCY VARIATIONS

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ABSTRACT

The relationship between pulse rate (PR) and pulse width (PW) for a constant level of electrocutaneous stimulation was ascertained using the method of comparative judgements. Twelve volunteer subjects were asked to adjust the pulse width of a Comparison Stimulus (S_2) set at 5, 10, 20, 50 or 100 pulses per second (pps) until S_2's intensity matched that of a Standard Stimulus (S_1) whose PW was 200 us and PR was 10 or 20 pps. As expected, the experimental results indicated that the PW of a constant curent amplitude pulse train should decrease as its PR increases if a constant level of tactile stimulation intensity is desired. However, PW and PR were not linear inversely related ($\underline{p} < 0.005$). Rather, their relationship was best described by a logarithmic equation: log PW = $\underline{a}$ + $\underline{b}$ log PR, where PW is in microseconds, $\underline{a}$ is 2.82, $\underline{b}$ is -0.412, and PR is between 2 and 100 pps. Utilization of this relationship during electrical stimulation of the skin sense will decouple the intensity component of the tactile sensation from its frequency component, thereby enhancing the potential comfort and clarity of this sensory communication interface.

INTRODUCTION

Electrotactile (or electrocutaneous) stimulation has demonstrated its usefulness in artificial arm sensory feedback, reading and mobility devices for the blind, augmentation for insensitive hands, and tactile vocoders for the deaf [4]. The use of electrocutaneous stimulation (i.e., the stimulation of the tactile sense using electrical impulses to evoke vibratory sensations) has also been evaluated as a signaling device for airplane pilots who face severe sensory (visual and auditory) demands. In short the tactile sense can serve as an alternate information input channel for selected assistive devices for the sensory impaired.

Using the tactile sense as an alternative sensory input channel has definite advantages and disadvantages. The advantages of the tactile sense for such applications include its availability and accessibility. The skin sense is usually not encumbered with continuous reception of fast changing, critical sensory information, and it can be accessed from a variety of body locations. The tactile sense, however, has two major drawbacks -- its limited dynamic range and information channel capacity for electrocutaneous signals.

Psychophysical studies [1,3] have indicated that the skin sense's dynamic range for electrical stimulation -- from perceptual threshold to pain threshold -- is about 10dB. The auditory sense, in contrast, has a dynamic range of about 140dB. Studies of skin discrimination characteristics have also indicated that 2-5 bits/sec is the maximum information rate obtainable using a single bipolar electrode as the input [4,5].

BACKGROUND

In order to fully utilize the 2-5 bits of information per second available from each electrode, investigators have modulated the frequency and/or intensity of the tactile sensations elicited at the stimulation site. Prior, Lyman, Case, and Scott [2] attempted to encode both grasp force and hand opening information in an artificial arm using a single electrode. Frequency variations via pulse rate modulation encoded hand opening while intensity variations via pulse width modulation encoded grasp force. They found this approach to be only partially successful because pulse rate changes produced concomitant intensity variations. Hence the amputee's perception of the two sensory feedback signals (hand opening and grasp force) from the prosthesis was confounded, resulting in cognition errors.

If the intensity component of the electrotactile sensation could be made independent of the frequency component, then each one could encode a different piece of information. For example, in mobility aids for the blind, the range of an impending obstacle might be transmitted using pulse rate modulation to control the frequency of the tactile sensation while the azimuth of the obstacle could be transmitted using pulse width modulation to control the intensity of the sensation. The desire to separate these two fundamental sensory components motivated the psychophysical study described in this report.

METHOD

Selection of Stimulus Parameters

The useful frequency range for electrocutaneous stimulation extends from about 2 to 100 pulses per second (pps) [3]. At a given frequency, the intensity of the electrotactile sensation depends on the amount of electric charge applied per pulse, which is proportional to the pulse width times the pulse amplitude. By using a constant current electrocutaneous stimulator capable of delivering up to 15 mA of peak pulse current, the entire dynamic range of stimulus intensity could be spanned by varying the pulse width (PW) between 30 and 1000 μsec. For these reasons, this study was conducted within these pulse rates and pulse widths.

In order to logarithmically divide the frequency range into approximately five equal parts, the pulse rate (PR) of the Comparison Stimulus was set at 5, 10, 20, 50, and 100 pulses per second (pps). The PR of the Standard Stimulus was set at 10 and 20 pps with a constant pulse width of 200 μsec. These parameters appear to be where humans exhibit the smallest just noticeable difference or the greatest differential sensitivity to electrocutaneous signals.

Experimental Procedures

Twelve students (6 male and 6 female), responding to an advertisement, participated in this study to ascertain the apparent relationship between pulse rate and pulse width for a constant intensity level of electrocutaneous stimulation. The method of comparative judgments was used to establish the pulse width adjustments necessary to make a Comparison Stimulus (S_2) equal in intensity to a Standard Stimulus (S_1). These two stimulus trains, both set at the same pulse amplitude, were alternatively applied to a subject's left forearm via a silver concentric bipolar electrode. His other arm adjusted the pulse width of S_2 up or down until the two pulse trains appeared to have the same intensity level. To prevent possible deleterious skin effects due to the stimulation, cathodic monophasic pulses from the stimulator were capacitively coupled to the electrode and the subject's forearm was premoistened with water. This type of electrode and stimulus waveform has been shown to be well tolerated by the skin [4].

To counterbalance possible persistence effects, such as errors of habituation and anticipation, three ascending and three descending measurements were taken in random order at each test frequency. For ascending measurements, the PW of the Comparison Stimulus was initially set to yield an intensity level clearly below that of the Standard Stimulus. Using his/her nonstimulated arm, the subject rotated a large PW control knob in the clockwise direction, causing the PW and thus the intensity of the Comparison Stimulus (S_2) to increase. When the subject judged the two alternating pulse trains, S_1 and S_2, to be equal in intensity, he stopped turning the control knob, and the experimenter recorded the pulse width of the Comparison Stimulus.

For descending measurements, the output pulse width of stimulator #2 was set at an initial value such that S_2 was clearly more intense than S_1. The subject then slowly turned the control knob counterclockwise, decreasing the PW of S_2, until the tactual sensations elicited by it equaled the intensity of S_1. At this point the experimenter again recorded the chosen pulse width.

The test apparatus consisted of the following: two identical electrocutaneous stimulators which produced the Standard and Comparison Stimuli; a specially constructed subject control box which held the circuitry needed to vary the pulse width of S_2; and an electronic switch which alternately connected the outputs of stimulators 1 and 2 to the cathode part of the bipolar concentric electrode or to the bypass resistor. The bypass resistor approximated the impedence of the premoistened skin and prevented unnecessary electrotactile stimulation of the subject during periods of adjustment and set-up.

Pulses delivered to the skin surface were measured using a precision 100 ohm current sensing resistor in conjunction with an oscilloscope and counter. These two test instruments, both hidden from the subject's view, enabled the experimenter to catch possible artifacts of stimulation, malfunctions of the test apparatus, and decidedly false responses of the subject. To maintain alertness and prevent possible sensory adaptation, the subject was also offered periodic rests from sitting and stimulation during the 90 minute test session. To give the subject enough time to feel and compare the two stimulus pulse trains, the electronic switch alternated between S_1 and S_2 once every 2-3 seconds.

ANALYSIS OF RESULTS

The pulse width compensation data for matching the Comparison Stimulus at 5, 10, 20, 50, and 100 pps to the Standard Stimulus set at 10 pps and 200 µsec or 20 pps and 200 µsec were plotted on rectilinear graph paper (Figure 1). In order to maintain a constant level of stimulation intensity, the pulse width should decrease by a factor of 3.5 as the pulse rate of the electrocutaneous signal increased from 5 to 100 pps. Furthermore, the relationship between PR and PW for a constant level of stimulus intensity is not linear in the least squares sense ($F(3,45) = 16.71$, $p < 0.001$ for S_1 at 10 pps and $F(3,55) = 5.95$, $p < 0.005$ for S_1 at 20 pps).

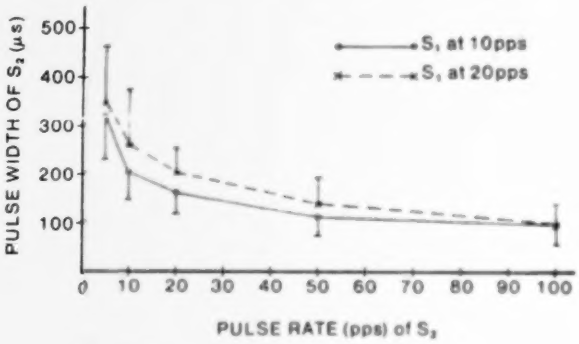

FIGURE 1. PW versus PR on linear axes.

Since PR and PW were not linearly related, the PW compensation data were then plotted on semilogarithmic paper (PW vs. log PR), shown in Figure 2, and on log-log paper (log PW vs. log PR), shown in Figure 3. Linear regression analysis of the PW data for semilog and log-log relationships revealed that the latter relationship accounted for more of the variance than the semilog relationship regardless whether the standard stimulus was 10 pps or 20 pps.

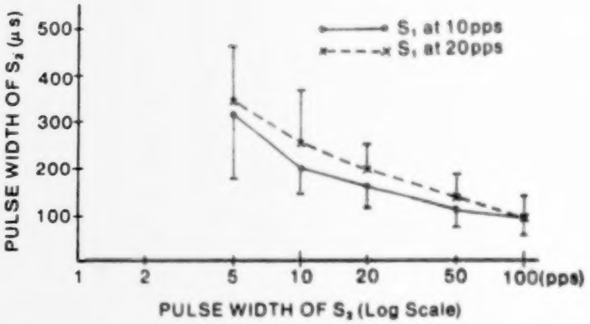

FIGURE 2. PW versus PR on semilogarithmic axes.

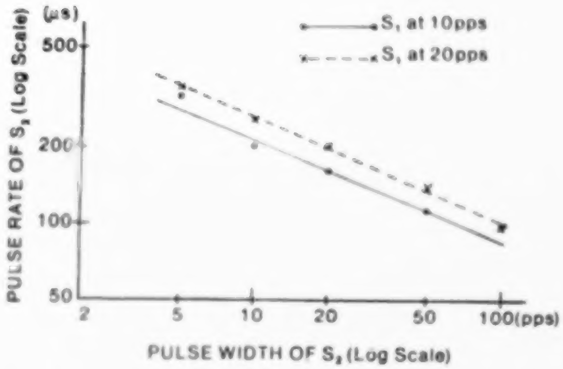

FIGURE 3. PW versus PR on log-log axes with linear regression lines plotted.

The regression lines for PW versus log PR matched to within 1%, but the regression lines for log PW versus log PR were slightly different. The slope and log PW intercept of the regression line for the 20 pps data were, respectively, steeper and larger than the slope and log PW intercept of the 10 pps regression line. However, the difference in slopes was not statistically significant at the 0.05 level. The difference between the log PW intercepts for S_1 at 10 pps and 20 pps was attributed to the fact that the first set of PW compensation data was adjusted to achieve an exact PW match between S_1 and S_2 at 10 pps whereas the second set of data were adjusted to achieve an exact PW match at 20 pps.

DISCUSSION

In summary, the above results indicate that the pulse width of a constant current amplitude pulse train must decrease as its pulse rate increases if a constant level of stimulation intensity is desired. However, these two parameters, PW and PR, are not linear inversely related. Rather, their relationship is best modeled by the following logarithmic equation: $\log PW = \underline{a} + \underline{b} \log PR$, where PW is in microseconds, $\underline{a}$ is $\underline{2.82}$, $\underline{b}$ is -0.412, and PR is between 2 and $\underline{100}$ pps, the practical frequency range for tactile communications [4].

By incorporating the above logarithmic relationship between PR and PW into an electrocutaneous communication system, one could imbed the informational signal into frequency variations while avoiding changes in intensity of the tactile sensation. Doing so would likely enhance the comfort and clarity of the tactile sensation and hence the efficacy of the sensory communication aid.

If one wished to maximize the information input rate achievable using a single bipolar electrode, then one could allow a second informational signal to vary the pulse width of the electrocutaneous signal above and below that needed to maintain a constant level of stimulation intensity. In this way, two channels of information could be transmitted to a person using both intensity and frequency cues.

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13.5

USE OF SOLID-STATE IMAGING COMPONENTS IN A READING AID FOR THE BLIND

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ABSTRACT

At the VA Rehabilitation R&D Center in Palo Alto, CA, work is underway to develop a compact and versatile reading aid for visually impaired persons.[1] The aid is to use fiber optics and a solid-state imaging device in its front end, an advanced 16-bit microprocessor to handle image processing tasks, and a custom interface board to allow it to drive specialized displays already in use by the target population.[2] The current paper describes components developed for incorporation into the device's front end "camera", which captures data for subsequent image processing and optical character recognition.

INTRODUCTION

It is estimated that there are approximately 500,000 totally blind persons in the United States, of whom perhaps 50,000 are veterans; additionally, there are some 1.5 million partially sighted but legally blind individuals, including over 200,000 veterans.[3] Many of these people could benefit from a device which would enable them to access inkprint material without the intervention of a sighted person, which would be portable and relatively inexpensive, and could drive existing displays already in their possession (e.g., an Optacon or stereotoner), and which could drive a speech synthesizer for voice output.

The objective of this project is to develop the prototype of such a device. In order to make the device compact, a small hand-guided camera will be used for image input. The blind user rests the camera directly on the page and guides it along lines of text, receiving either tactual or audio feedback to assist in the task. For processing purposes, camera images can be thought of as a series of overlapping snapshots of the text. Individual characters, words and text lines must all be reconstructed from these snapshots. Successful reconstruction will require accurate positional information of each snapshot, because there will be significant variations in the velocity and direction of the camera. This paper focuses on the "front end" portion of this project; thus, the imaging device, control of the imaging device, optics of the hand camera, position extraction and image reconstruction are all discussed.

THE OPTIC RAM

The imaging device which we are using in the prototype reading aid is an Optic RAM chip, manufactured by Micron Technology in Boise, Idaho.[4] This is essentially a 64K dynamic Random Access Memory (RAM) with the ceramic lid replaced by a window to allow the silicon chip (die) to be exposed to light. The device provides two sensor arrays, each containing 128 x 256 picture elements, or "pixels". Such a solid-state imaging device presents several highly attractive features: it is inexpensive (approx. $15 apiece in quantity, $40 singly), compact, durable, operationally fast, of high resolution, and able to alter its imaging window under software control.[5] Such features make it well suited for incorporation in our prototype reading aid. The two disadvantages of the Optic RAM appear to be tractable, namely that it requires frequent and uninterrupted refreshing in order to deliver accurate image data, and that it is capable of delivering only thresholded image data (i.e., not grayscale data).

PHASE-1 OPTIC RAM INTERFACE

At the RR&D Center, we have constructed a simple interface between the Optic RAM and a MC68000 microprocessor, in order to verify the suitability of the Optic RAM for our application. A sample image obtained using the Optic RAM and our Phase-1 interface board is shown in Fig. 1.

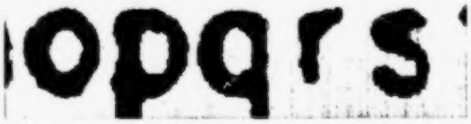

FIG. 1

Fig. 2 shows a block diagram of our Phase-1 interface board. Our board is a modified version of an interface board originally described in popular literature.[6] As the diagram shows, it operates entirely under control of the MC68000 microprocessor. The microprocessor can read bits from the sensor array by sending an address to the interface board, which translates the bit pattern into sensor location coordinates and formats them for receipt by the Optic RAM. The Optic RAM then sends back the datum from the specified location.

An important aspect of the control of the Optic RAM is the periodic "refreshing" of its imaging cells. Refreshing is required in all dynamic memory devices to prevent loss of data. In the Optic RAM, cells on any row may be refreshed by reading any cell from the same row. Therefore, adequate refreshing can be accomplished during a sequential read-out of the array by reading the cells in a column-by-column fashion. In this way, each read accesses and refreshes a different row, and the entire array is refreshed every 128 reads.

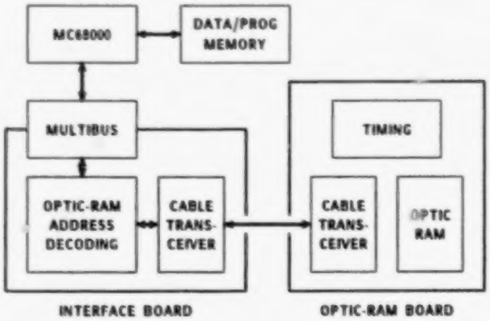

FIG. 2

Experience with the Phase-1 interface board convinced us that it would be more efficient to perform many of the repetitive "housekeeping" tasks associated with the control of the Optic RAM in hardware rather than under microprocessor control. To this end, we have designed a more intelligent, Phase-2 version of the Optic RAM interface.

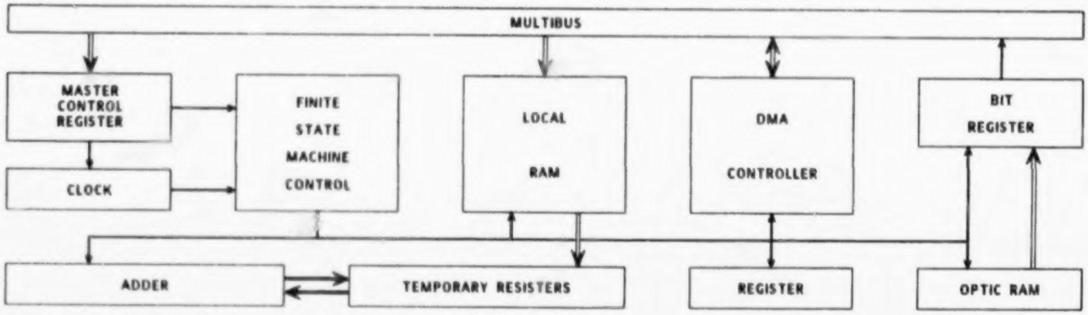

FIG. 3

PHASE-2 INTERFACE BOARD

Fig. 3 shows a block diagram of our Phase-2 interface board, which is now under construction. This board will perform many Optic RAM control and interface functions in hardware, freeing the microprocessor for other tasks. For one thing, it will handle all Optic RAM housekeeping and refreshing tasks automatically. In addition, it will simplify pixel addressing by translating (x, y) geometrical coordinates into the Optic RAM's peculiar internal addressing scheme. Moreover, it will allow the microprocessor to specify the locations and sizes of "windows" within the sensor array, and automatically transfer the contents of these windows to the microprocessor's memory via direct memory access (DMA). The DMA transfer will place groups of 16 vertically or horizontally aligned pixels into each word of microprocessor memory. These features will make the operation of the Optic RAM essentially transparent to the microprocessor.

Eventually we intend to include all Optic RAM control and interface functions on a single custom integrated circuit. We expect that our experience with the Phase-2 board will enable us to specify the precise functions needed for our application.

PROJECTED CAMERA CONSTRUCTION

Hand-held Camera

It is our intent to integrate the Optic RAM and interface into a compact hand-held camera. Our goal is to produce an efficient and reliable camera, which will perform all housekeeping tasks in a way which is transparent to the main microprocessor. In this task, we would like to employ inexpensive contemporary technology in an novel fashion, in order to produce a unit which is compact, rugged, versatile and affordable. In this context, three additional matters deserve mention: the optical system, our approach to miniaturization, and techniques for estimating position and velocity.

Fiber Optics

An Optic RAM can deliver good image data only if there is a clear image focused upon it in the first place. This, in turn, requires the use of optical components. For this particular application, we are looking at using Selfoc fiber optics lens arrays[7]. These units, manufactured in Japan, are most commonly found in copying machines, where they deliver a long narrow image from one plane (that of the paper to be copied) to another (that of the copying drum). The lens arrays employ a staggered line-up of relatively large diameter fiber optic cylinders, each with a variable index of refraction, that generate overlapping images on the imaging plane. The cylinders are so arranged that the overlapping portions of the images reinforce one another, producing at the imaging plane an image of high quality. Normally these lenses come in strips around 10 inches long and a few millimeters wide; for our purposes, a length of approximately half an inch should fill the Optic RAM's imaging area. Since the Selfoc lenses do not need to be physically in contact with the inkprint being imaged, they can float a proper distance above the page, allowing space for illumination of the text below.

Miniaturization

We would like to produce a compact and relatively inexpensive hand-held imaging unit. In our current constuction iteration, which uses printed circuit boards and TTL medium scale integrated chips, the Optic RAM controller is large and expensive, and must reside in a separate chassis from the Optic RAM. In order to produce a truly compact hand scanner, we intend to package the controller onto a single custom integrated circuit (IC) located in the hand scanner itself along with the Optic RAM and the illumination system. The board depicted in Fig. 3 above is meant to clarify precisely which functions are to be included in that IC. The custom IC will be designed and fabricated at the Stanford University IC laboratory.

Position and Velocity Detection

Software is being developed and tested which will estimate position and velocity directly from data received from the Optic RAM. The approach is to calculate the correlation between successive frames to determine how far and in what direction the camera has moved since the last frame. For these purposes, we are planning to use two of the software controllable "windows" described above as special "correlation windows", whose data will be used solely for the purpose of detecting and reporting motion. This approach for obtaining positional data is attractive because it eliminates the need for a device which requires continuous mechanical coupling with the page, such as a shaft encoder or a resolver. The general approach has been shown effective in earlier work done at Telesensory Systems, Inc., using image data from an Optacon camera (a device conceptually similar to the Optic RAM, but with far lower resolution). For collecting image data for optical character recognition, another "window" will be used. These data, oriented using the positional information from the correlation, will be used for image reconstruction. Lines of text will thus consist of sequences of characters reconstructed from camera images.

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A Simplified Touch-Tone® Telecommunication Aid for Deaf and Hearing Impaired Individuals

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ABSTRACT- A device has been completed that allows a normal-hearing individual to communicate with a deaf or hearing-impaired person from a standard Touch-Tone telephone. The device, the T (Touch-Tone Text) Decoder, is an assemblage of hardware and software that converts Touch-Tone signals into readable text which is presented to the deaf person. The single-stroke text entry permitted with this system is twice as fast as competitive Touch-Tone methods. The device correctly decodes over 95% of running text. The T Decoder has proved more than adequate for initiating and maintaining telephone conversations (12).

INTRODUCTION

The deaf and hearing impaired lack an effective means to access the telephone network, especially for conversations with normal hearing individuals. The two widely employed methods for telephone use, namely interpreters and telecommunication devices for the deaf (TDDs), have severe drawbacks (2,3,8). The interpreter technique precludes privacy and requires that a normal-hearing person be available at all times. TDDs, while affording privacy and independence, are expensive and necessary on **both** ends of the line.

BACKGROUND

Touch-Tone dialing signals have been repeatedly proposed as a medium for transmitting messages to the deaf and hearing impaired with standard telephones (1,5,7,9,10,13,16,20). A normal hearing person would no longer be required to posses a technical aid. Instead, they would enter the text of their end of the conversation on the Touch-Tone keypad (Fig. 1). The device, residing with the deaf or hearing impaired person, decodes the Touch-Tone signals and visually presents text to the user. The deaf or hearing impaired person then verbally responds to the normal hearing person.

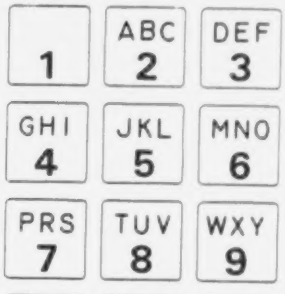

Figure 1. A diagram of the standard Touch-Tone keypad.

An important difference between the Touch-Tone keypad and a standard typewriter is the number of letters appearing on each key. Each previously proposed device has required that the normal hearing person strike at least two keys for each letter. Using the Multi-key Code, the sender first strikes the key on which the desired letter appears and then strikes the 1, 2, or 3 key to indicate which letter was intended (5,13,20). The second method, the Multi-strike Code requires as many as four keystrokes for each letter (10).

In contrast, the T Decoder permits a sender to simply type the key that the desired letter is on; no additional keystrokes are necessary (12). It was believed that the reduced number of keystrokes required would allow faster communication, the similarity to typing would speed learning times, and the simplicity would reduce error rates and sender dissatisfaction.

Twelve subjects were taught all three techniques to investigate these hypotheses. Using the single-stroke method, text was entered by these novice users at 11 w.p.m., nearly twice as fast as the competitive methods (Fig. 2). Subjects asked fewer questions about the rules reguarding single-stroke text entry (note that questions cannot be addressed to the deaf person during actual device use). Subjective preferrence for the single-stroke method was unanimous (12).

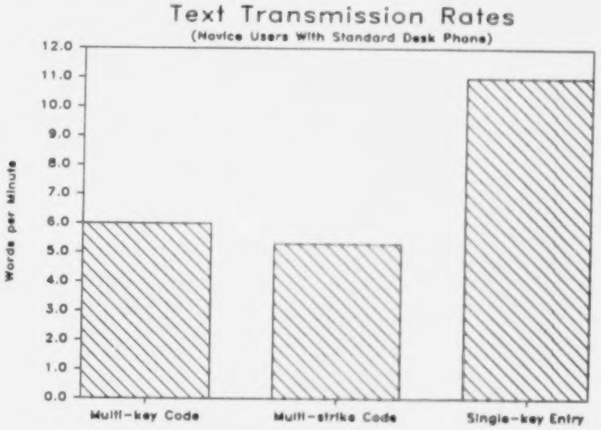

Figure 2. Results of Text Entry Experiment.

METHOD

The T Decoder was implemented using an IBM® PC™ and off-the-shelf accessories. The utilization of commercially available hardware was an important aspect in the design of the device. This allows the T Decoder to take advantage of the economies of scale, warranties, and service networks of the component's manufacturer. Also, with the hardware to perform Touch-Tone keystroke extraction no longer a major element in the device design, additional effort could be directed towards the user interfaces and intelligent interpretation of the incoming keystrokes.

Two specific Touch-Tone decoding devices have been employed in the implementation of the design; each has features applicable to the needs of particular individuals. The Magnum-10™ board is an economical board to perform the functions of Touch-Tone decoding and acoustic modem. The DECtalk™, on the other hand, extends the use of Touch-Tone communication devices to that portion of the deaf and hearing population lacking intelligible speech. That is, the deaf or hearing impaired user would still read from the T Decoder but would now reply by typing their response to the DECtalk, a high-quality speech synthesizer.

The heart of the T Decoder is the software. Previous Touch-Tone devices have each required a code that established a reversible deterministic relationship between the string of digits being transmitted and the text it represented. With the T Decoder this relationship has been sacrificed in the interests of sender convenience. Where a previously proposed device might receive the digits 8, 2 and know that the letter U was intended, the T Decoder receives only an 8 and knows it to represent **either** a T, U, or V. The T Decoder's software converts a string of digits, each representing one of three possible letters, into a readable text string.

This is accomplished with a hybrid algorithm, two approaches to the problem running in parallel (17). The results from this hybrid is considerably superior to what either technique could accomplish alone.

One of the techniques is based on the probability of each possible string of letters. For instance, if the string •843• is received (• represents the character **space**), we know that the intended word must be one of the 27 possibilities given in Figure 3. These possible words are scored using the product of their component trigram probabilities (14,18,19). The top scoring strings in the example are marked in the figure.

TGD	TGE	TGF	THD	THE-1	THF	TID	TIE-4	TIF-3
UGD	UGE-6	UGF	UHD	UHE	UHF	UID	UIE-5	UIF
VGD	VGE	VGF	VHD	VHE	VHF	VID-2	VIE	VIF

Figure 3. Possible Words for Digit String •843•.

The other approach involves looking in a dictionary for a word that can be represented by the received digit string. The frequency distribution of English speech reveals that a small number of words comprise a large proportion of speech (4). Analysis of the Touch-Tone strings for these common words shows that more than 90% of these words have unique codes (12). A dictionary of these words can be searched upon receiving a Touch-Tone string; if a word is found, there is a high probability that it is the intended word (14).

In addition, dictionaries customization of the device for recognition of words specific to a user. For instance, the names of people commonly called or activities frequently discussed can be added to the dictionary. This will make the output from the device easier to read.

RESULTS-

The hybrid algorithm has been evaluated in two ways. One method involved transforming text files into corresponding Touch-Tone form, inputting the resultant digit string into the program and comparing the output text to the original. The algorithm correctly spells greater than 96% of the input text! Detailed results of this procedure are shown in Figure 4.

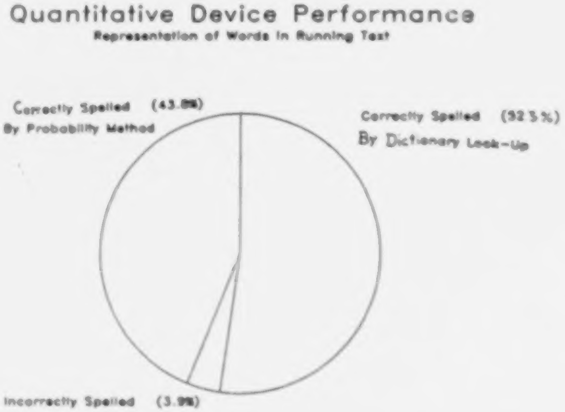

Figure 4. Results of Text Decoding Task.

In addition, trial conversations have been conducted using the device. These have been completely successful; the context of a conversation makes it simple to interpret the occasional poorly represented word.

CONCLUSIONS

The T Decoder, as presented, is a viable communication device that offers a number of improvements over any currently available aid. The minimum cost of a working system is approximately $2400. In evaluating the cost of this system, it should be noted that it also allows access to the DEAFNET network via its modem and that the IBM PC is fully functional for running other software when a phone call is not in progress.

The T Decoder will require a minimal committment to manufacture and distribute. It is an assemblage of commercially available hardware and a single diskette of software. Service, if necessary, would be available from the original manufacturer of the component.

The next breakthrough for the T Decoder will be its implementation on a portable computer. The software, with some alterations, will run on a portable, "notebook" computer. This will allow the deaf and hearing-impaired to place and receive calls from any location, even a phone booth! The cost is of these computers is quite reasonable, the Portable T Decoder could conceivably sell for less than $600.

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14.1

THE R.S.O.:
A Recumbent Support Orthosis for Control of Lower Limb Deformities in Children with Neuro-muscular Disabilties

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ABSTRACT

Children with the more severe grades of Cerebral Palsy and Myelomeningocele have strong tendency to develop knee and hip deformities. We have observed that these deformities develop in accordance with the child's unusual and consistent sleeping posture. During the past seven years we have developed a rather simple, adjustable orthosis for night-time use which supports the hip and knees in a preferred alignment while the child is sleeping.

* * *

Children with spina bifida, spinal cord trauma, or the more severe grades of cerebral palsy, have a strong propensity for the development of progressive hip and knee deformities. Those deformities usually have their origin in pre-existing muscle imbalances which pull the legs into positions which are consistent and repeated each nap and nighttime. Gravity and biomechanical factors seem to work together in a mischievous way.

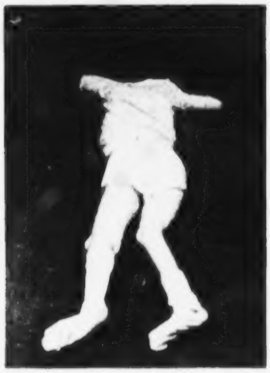

Figure 1

For instance: knee flexion contractures alone, under the effect of gravity, will cause the hips to be rotated (especially when prone) and flexed when the child is recumbent on a flat surface. Hip flexion contractures cause a recumbent position which includes hip rotation of either the frog-leg or the wind-blown type. There is no one age when these deformities develop. In some children, they appear in infancy; others begin significant, progressive knee and hip deformities as they enter their second decade and spend less time upright and more time sitting.

Anytime a child spends eight or more hours per day, every day, in a consistent position of paralytic origin, the child will probably develop joint contractures and those contractures will be progressive. Significant knee and/or hip contractures will cause ambulation or standing to be more risky, more difficult, and more energy costly, if not impossible. When the deformities are asymmetrical, they cause poor sitting postures. They also make it more difficult to avoid the generation of pressure sores during recumbency.

The use of sand bags and pillows to achieve an acceptable sleeping body alignment usually does not work because it means, in effect, rebuilding a supporting structure every night in a way that appreciates:

1. The geometry of deformities,
2. The biomechanics of simultaneously resisting those deformities, and
3. The need to avoid excessive pressure areas.

In our experience, the sand bag and pillow approach is inefficient and in the long-run just does not work.

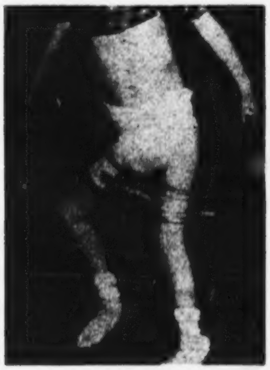

Figure 2

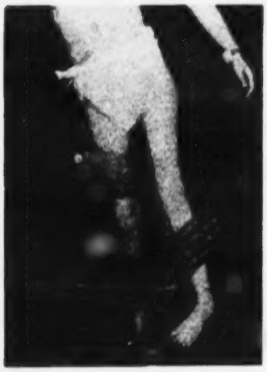

Figure 3

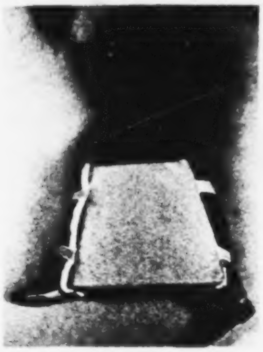

Figure 4

Such things as adductor cuffs which are used to resist bilateral hip abduction (frog-leg) and abduction wedges to resist hip adduction have been dismal failures for two reasons;

1. They do not cross superior to the hip joints and therefore cannot be expected to control hip position; and
2. The contracture or spasticity will always be stronger on one side so any deformity reduction will occur only on the least severe side.

In figures 2 and 4, you can see exactly why the adductor cuffs or the abduction pillow does not effectively hold both hips in abduction. Bilateral abduction is usually not the position achieved. The child can very easily assume a position of unilateral abduction or unilateral flexion with no abduction. Also, of course, they have very little constructive effect on hip rotational position.

As an illustration, the child you see in figure 2 underwent surgery three years ago. That surgery was meant to correct bilateral abduction contractures. Following surgery, this adduction cuff device was prescribed. As you can see, the abduction deformity on her right side reoccurred and dominated. There has been no benefit. The surgery and treatment has merely traded one deformity (the frog-leg) for another (the windblown). Figure 3 shows that a neutral position, however, is achievable if the correct set of support forces is used.

About 1978, we started to fabricate and provide devices designed to resist, during recumbency hours, the development and progression of hip and/or knee contractures. We call the devices Recumbent Support Orthoses because they support the child in a preferred alignment while recumbent.

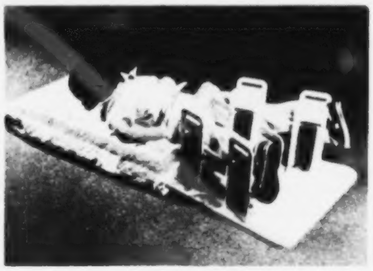

Figure 5

The basic RSO design is adaptable to any combination and severity of lower limb deformity and allows considerable growth adjustment. Since it is not form-fitting, it is relatively less expensive. It accomplishes its goals at night, when the child is in bed sleeping and therefore does not interfere with daytime activities.

The RSO can involve various design configurations. The RSO design can hold one hip in neutral and one hip in abduction with both in full extension.

Figure 6

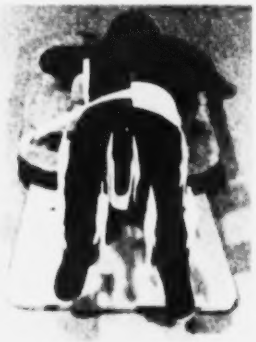

Figure 7

Figure 6 displays a windblown deformity which was treated with an RSO design shown in figure 7. To achieve hip rotational control, the RSO can hold the child's hips in neutral and full extension, with a bar attached across shoes to prevent rotation.

The orthosis is also regularly used at G.C.H. as a means of post-operative positioning following surgical release of hip and knee contractures. As you perhaps know, paralytic hip and knee deformities will reoccur following surgical release unless a long term positioning program is faithfully adhered to. At Gillette we have found a period of at least one year is required of a successful positioning program following these surgical procedures.

The "Gillette" Recumbent Support Orthosis can be designed for either the supine or the prone recumbent position. It can also be designed, if it is holding a symmetrical position, so that the prone and supine positions can be used on alternate nights or be at the option of the child or parent. This reversible design, though, requires some strap changes which not all care-givers can accomplish.

Education of the care-givers in the proper usage of this orthosis is very important. They must have a solid understanding of how the orthosis works. Some parents with good initiative and excellent understanding of the orthosis can adjust it to achieve a significant all-night stretch of contractures. Realistically we are not achieving that in most cases. However, we do achieve maintenance of a more neutral joint alignment which does not reinforce the contracture.

Originally, we thought cosmesis not to be a problem since the use of the RSO is confined (usually) to the child's bedroom. However, we discovered that some parents resisted its use. We think that was partly because we were depriving them of the gentle, loving act of tucking their child into their soft bed. Figure 8 shows design changes occurring since we have recently paid more attention to its appearance and are trying to make it appear a bit softer and more attractive as a place to lie your child down for the night.

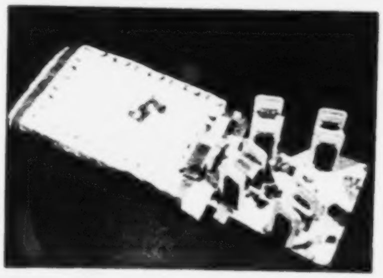

Figure 8

Since 1980, the Recumbent Support Orthoses at GCH are fabricated and fit by David Wilkie and others in his Adaptive Equipment team in the Orthotic & Prosthetic department. David is responsible for many design improvements which have made the orthosis more adaptable, more cosmetic, lighter, and less expensive.

The personnel of the Rehabilitation Therapies Department at Gillette (Karen Beck, RPT, in particular) have been important in the development of the application criteria for the Recumbent Support Orthosis in our client population.

If you have any questions regarding use and design of the RSO please contact the author at the following address:

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PROTECTIVE HEAD GEAR: THE GILLETTE CHILDREN'S HOSPITAL APPROACH

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ABSTRACT

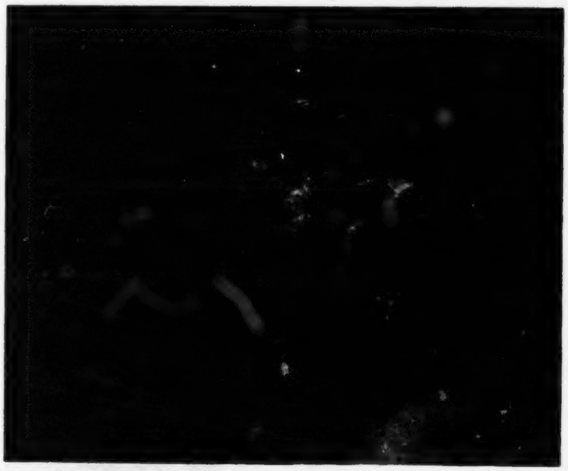
This paper deals with providing protective head gear for patients whose heads are at risk of injuries. Commercially available helmets and custom fabricated helmets have both been used successfully for various disorders at Gillette Children's Hospital.

The adaptive equipment area at Gillette Children's Hospital is involved in providing head protection for many clients with various disorders. The bulk of the client population whose heads are at risk are those who fall readily, engage in self abusive behavior or whose skulls are partially absent. Providing head protection for these clients can be difficult and frustrating, yet when successful, can be extremely rewarding. At Gillette we have developed an approach that works well in our area with our client population.

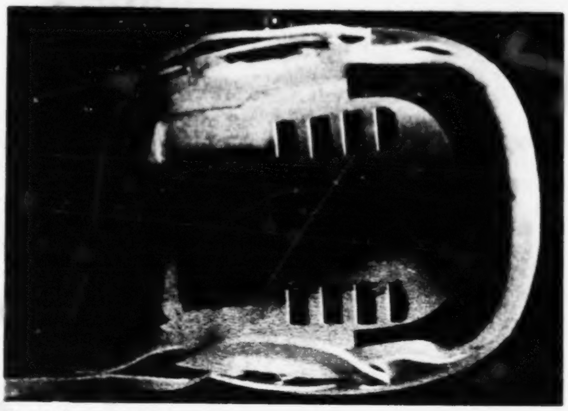
Choosing a helmet design to fit the needs of the client is the first step. The helmet that is chosen must offer maximum head protection, minimum head coverage, and good ventilation. It must be lightweight and as cosmetically acceptable as possible.

The appearance of a helmet can be very disturbing to the client, parents or caretakers. Acceptance of the helmet may be difficult and many times the decision to use the helmet is based on a "lesser of two evils" approach. Helmets are hard, if not impossible, to conceal and the best approach is usually not to try to conceal them. A bonnet type covering may be used over the helmet of small children successfully. Some clients have elected to use decorative painting or colorful stickers on their helmets.

We recommend the use of commercially available hockey helmets whenever possible. These helmets come in many styles and colors and most are adjustable in size. The adjustability makes fitting easier and allows for considerable growth. The adjustability also allows for a close fit, making the helmet look less obtrusive. With modifications to the interior padding and special strapping, hockey helmets are being used for 85% of our clients.


Craniectomy client

Commercially available hockey helmet


Padding added to transfer pressure

Karate helmets have been used for clients who continually bang their heads, but these helmet are quite hot, cosmetically unacceptable and not very durable.

Hockey helmets may not be available for the very large client. In these limited cases we resort to the use of a baseball batters helmet. The "cap like" appearance of these helmets make them more readily accepted by parents and caretakers. They offer very good protection and a hockey face mask can be attached. They are non-adjustable, making fitting more difficult, and must be ventilated.

When deciding on a helmet design we should first decide if the client requires face protection. This is extremely important because it determines the material and strength of the helmet. This decision may be made by investigating seizure activity and previous injury locations. If the client exhibits patterns of injury only to the forehead, back of the head or sides of the head, face protection most likely will not be required. Many times the epilepsy client will exhibit these injury patterns resulting from their seizures. Clients who do not require face protection can be fit with lighter, more cosmetic and less expensive head protection because a face mask does not need to be attached.

At Gillette, we custom fabricate protective head gear only as a last resort. Custom made helmets are used in all of our unusual cases such as hydrocephalus, extremely deformed heads and for small children who require head shaping helmets. These helmets provide very effective protection, but are time consuming and expensive. Model taking and modifications for these helmets can be difficult. At Gillette we use a bean bag dilatancy casting approach to obtain our model whenever possible. This approach may not be practical with craniectomy clients or some other special clients. For these clients, precise measurements can be taken and a model can be carved out of plaster, but this requires a skilled craftsman who many times is not available. Commercially available helmets can sometimes be modified in these cases.

Helmets for craniectomy clients can be extremely difficult to fit. After removal of portions of the cranium, the symmetry of the head may be lost and the bone area necessary for distribution of pressure over the surface of the cranium, may be drastically reduced. In most craniectomy clients, it is important to transfer the helmet weight and pressure to the existing forehead, size of the head and occipital regions, leaving the craniectomy site with little or no contact. Prevention of helmet migration in all types of helmets is extremely important and can be accomplished by the addition of carefully placed interior padding and special chin straps.

Clients who exhibit self-abusive behavior to their face and head can be extremely challenging. Many of these clients are able to carry on functional activities with their hands, making any type of restraints nearly impossible. These clients must be provided with helmets which will not migrate on their heads, allowing protection to remain in the desired location. A very high percentage of self-abusive clients will attempt to remove any head protection provided. Effective means of strapping these helmets must be used along with fasteners that are both difficult to open and difficult to reach for the client, yet manageable for the caretakers. In order to stop this type of client from reaching his face it may be necessary to provide almost total face coverage. In the persistent client, this protection may have to extend to the anterior neck and chest areas to prevent the client from going under or around a normal face mask. Padding of the outside of the helmet may also be necessary if the client continues to attempt to hit his face. Refitting of these clients on an ongoing basis may be required if their behavior changes.

Helmets in our society are becoming more and more visible. Bicyclists, motorcyclists, sports figures and astronauts to name a few, are visible in our society. Hopefully, this will have a positive effect on the acceptance of protective head gear in the future.

Acknowledgment: Some of the protective headgear provided at Gillette is designed and fabricated by Lee Hegfors and John Spielman, Adaptive Equipment Specialists.

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DEVICES FOR SUPPORT AND TRANSPORTATION OF THE ACUTELY SPINAL INJURED: CLINICAL TRIAL AND DESIGN ITERATION

14.3

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* Dept. of Physical Medicine & Rehabilitation,
Santa Clara Valley Medical Center

ABSTRACT

This paper presents results of field trial of a constant-force-spring traction device and folding laminated wood backboard [1] for transport of acute spinal injury patients, and progress of advanced designs incorporating these findings and experience with an intermediate prototype [2]. After approximately 60 admissions via helicopter to Santa Clara Valley Medical Center over 2 1/2 years, plus intra-hospital transport for computerized tomography (CT) and conventional X-ray, cumulative damage to exposed springs compromised the nominal spring output, but the device remained functional due to inherent redundancy. The new design, a carbon-fiber/epoxy foam-core composite backboard and cable-output constant-force traction has been released for trial following bending tests of composite specimens.

OBJECTIVES

The goal of this project is to reduce the risk of further trauma to cervical spinal injury patients by design of stabilization and patient maintenance devices compatible with inter- and intra-hospital transport equipment, CT [3] and nuclear magnetic resonance (NMR) [4] scanners, conventional radiography and hyperbaric oxygen (HBO) chambers.

The backboard is not intended to be an extrication device, although adaptation of components for extrication is possible, but rather to be a means for applying traction immediately after primary treatment and initial diagnosis of spinal injury, and to maintain it uninterrupted during inter- and intra-hospital patient movement.

This development phase is concerned with: (1) improvement of lateral head restraints to increase rigidity, eliminate interference with traction fixtures and reduce complexity; (2) combination of the best features of previous prototypes; (3) replication of the backboard in carbon fiber/epoxy for increased strength and x-ray transparency; (4) reduction and simplification of effort required of flight and rehabilitation nursing staff; and (5) acquisition of data on physical properties, on in-flight dynamics and on effects on patient welfare.

BACKGROUND

Management of acute spinal trauma was reviewed by Albin, et al [5], and by de la Torre [6]. The problems entailed in transport of the acutely injured are of more general concern as aircraft increasingly become the method of choice [7,8]. Similarly, as CT scanning has become more common makeshift traction devices have proliferated [9]. Commercial development of devices specifically addressing problems of radiographic translucency and continuation of traction during diagnosis and therapy has been limited to the recently-marketed "Sukoff device" [10], intended for HBO but with optional torsion-spring traction for use in patient transport.

FIRST PROTOTYPE

This backboard was 18 inches wide x 81 long, with locking hinges at mid-length, of uniform 3/8 inch thickness and radius of curvature of 30 inches. Variation in head width was accommodated by lateral boards hinged to the torso board parallel to the body axis. After placing the patient on the board, these restraints were attached by threaded ball-end studs & knobs to the bracket above the head supporting the traction device; a similar clamp fixed small shoulder boards to the main board (Figure 1). Fabric-covered "Sun-mate" foam pads were attached to the head and shoulder restraints; the torso pad consisted of a "Flo-lite" viscoelastic fluid-filled bag over 1 inch thick urethane foam.

The traction device [1] consisted of 9 "Neg'ator" springs on a common shaft. Pairs of springs were extended from the housing and attached to a summing bar using wingnuts, for increments of 5.2 lbf.; the center spring provided 2.6 lbf increments. An S-hook connected the summing bar to the halo or tongs.

Results

Laboratory tests were: (1) evaluation of traction force with various combinations of springs, using a spring scale, (2) test of restraint straps and head restraints with a volunteer subject in halter traction (5.2 lbf), (3) weights of components, and (4) X-ray opacity.

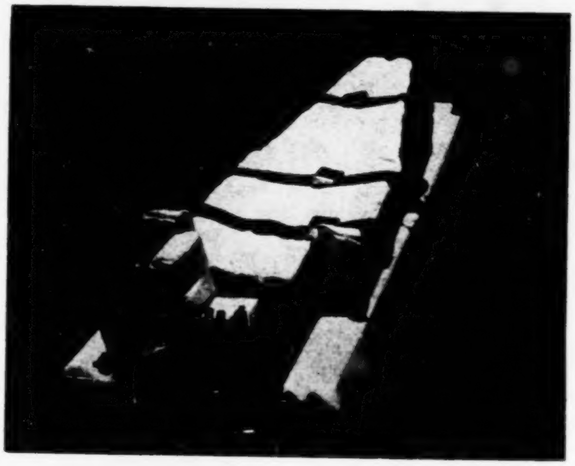

Figure 1: FIRST BACKBOARD-TRACTION DEVICE

The board as used for all admissions was ordinarily placed on the upper part of an old-type Stryker turning frame. Two hours stay on the board was typical. There were no reported problems with patient restraint during tilting for relief of respiratory distress. An elevating gurney was helpful for sliding the patient/board from the Stryker to a bed. The folding feature was not used.

The board was used routinely for CT scan, averaging one hour duration every other week, and occasionally by the Intensive Care Unit. The board was usually placed on the back of a prone patient in a Stryker frame, then turned to the supine position. Head restraints were seldom used, due to interference with halo pins. An adaptor for mounting the traction assembly on a turning frame was used once, found inconvenient, then lost.

On inspection of the traction device after 2 years use without maintenance, aluminum wear particles were found embedded in the spools. Six springs had no significant damage (ripple of <2 inch from free end); 3 springs had crimps & ripple of 4-6 inches. The shoulder boards & clamps had been removed as unnecessary. Pad covers had been lost in the laundry. The "Flo-lite"/urethane pad was in use and thought by staff to be an improvement over existing alternatives, although hard to keep clean.

ADVANCED DESIGN

The backboard (Figure 2) is 79 inches long and 5/8 inch thick, but otherwise similar to the previous prototype. It is fabricated by hand lay-up of skins of two orthogonal layers of "Orcoweb" G-450 graphite fiber in Epon 828/Versamid 140 (2:1) resin over a "Klegecell" 55-PL PVC foam core. The lay-up technique produces molded-in handholds and finished edges, and was designed to be a low-cost method for pilot-scale production.

Previous head restraints did not satisfy the requirement that restriction of lateral head movements not interfere with maintenance of traction force magnitude. Ideally, the restraint should be frictionless parallel to the cephalad/caudal axis, while being fixed parallel and lateral to the sagittal plane. This is accomplished by supporting the head and neck in a semi-rigid cradle sliding on tubular cantilever shafts held by a frame on either side of the patient's head (Figure 3). The cradle material is "Hexcelite" orthopedic casting mesh, a low-temperature thermoplastic coating on cotton yarn. The cradle is pre-formed to median head dimensions, but can be individually fitted by warming with hot-water-soaked towels or a blow-dryer. Anterior motion of the patient's head is restrained by a forehead band attached to the cradle cover by Velcro fasteners. The geometry of the cradle and head band leaves both the patient's ears and skeletal fixation pin sites unimpeded. The head restraint frame of Figure 3 is a preliminary version; the final design will have to eliminate interference with access by the attendant supporting the head while placing the patient on the board.

The new traction device retains 9 independent modules rated at 2.6 lbf each, but uses the cable output of the second prototype [2]. For minimum mass and volume, as well as redundancy, each module has two 10.4 oz-in "Neg'ator" B-motor springs acting on a single 1 inch diameter take-up spool/cable reel. The cables exit the housing via Teflon fairleads and are terminated by grip rings and pins, which insert into keyhole-shaped slots in the summing bar (Figure 4).

Mass of the backboard is 11 lb, the traction device 2.2 lb, and pads (fabric-backed vinyl over 1.5 inch medium "Sun-mate" foam) and straps 7 lb.

Testing

Samples of backboard composite were tested in 4-point bending at a rate of 0.5 mm/sec in a MTS testing machine (Figure 5). Ultimate stress, strain and energy absorption were consistent with published results for similar composites and adequate for the design static load of a 250 lb patient with a safety factor of 1.5, modeled as an end-supported point load with bending yield stress of 1350 N/sq. mm.

The traction cable output had significantly higher hysteresis (mean force difference between extending and rewinding) than springs from the first prototype (Table 1), due to friction of 3 spools per module (33% of available force) compared to one spool (18%) in the earlier version. Acircularity of the spring anchor and spool bores caused 16-33% higher oscillation during extension and retraction, but the design solved a monotonic force increase on extension noted in the first version. Friction and roughness can be reduced, but not eliminated; lubrication or ball bearings are undesireable if the assembly is to be used in HBO.

DISCUSSION

The first prototype was not capable of stand-alone function in interhospital transport. It suffered from weaknesses in (1) head restraint adjustability and interference with tongs and halos, (2) pad covering and attachment, (3) susceptibility of the constant-force springs to damage, (4) difficulty of attaching springs to the summing bar, and (5) insufficient quate off-axis angle adjustment. It has not yet been used for HBO or NMR treatment.

The new design is expected to have the same advantages as the earlier prototype, including simple, reproducible and redundant traction force magnitude, compatibility with other transport equipment and with CT scanners, and adequate acceptability by nursing staff. It is intended to be a component of a comprehensive system for diagnosis and treatment of cervical spinal injury within the few hours after trauma that reversal of neurological damage may be possible.

ACKNOWLEDGEMENTS

Supported by a grant from the Spinal Cord Research Foundation of the Paralyzed Veterans of America. We are grateful to Doug Schwandt and Rob Whalen of the PAVA RR&D Center and to Art Leach of Stanford University Mechanical Engineering Dept.

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Figure 2: GRAPHITE-EPOXY BACKBOARD

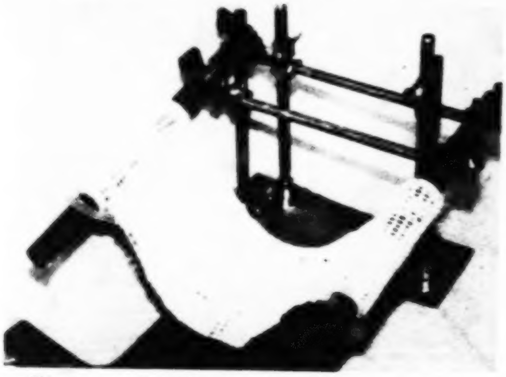

Figure 3: HEAD RESTRAINT TEST ARTICLE

Figure 4: CABLE-OUTPUT TRACTION DEVICE

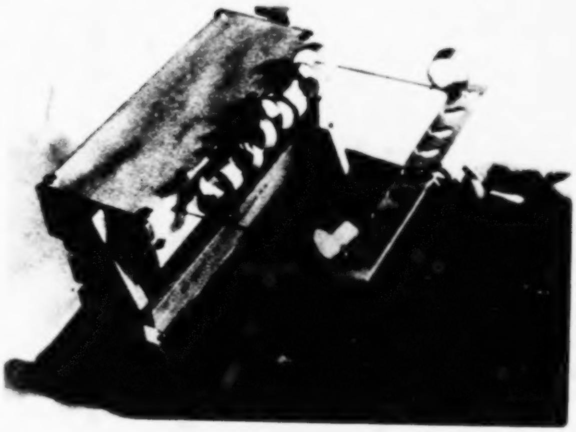

Figure 5: COMPOSITE SPECIMENS IN 4-POINT BENDING

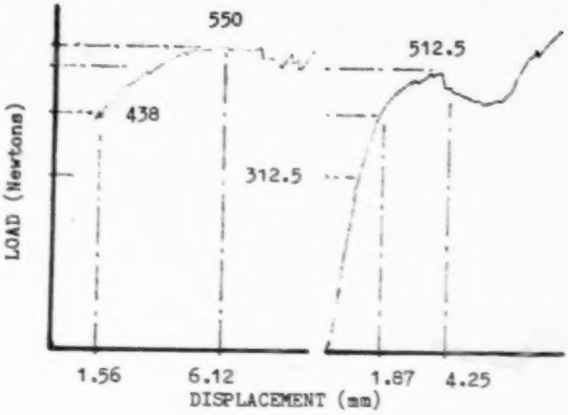

Table 1: TRACTION FORCE VARIATION

VERSION			FORCE (oz) MEAN	S.D.	N
1	extending	mean	52.4	4.0	18
		low	50.0	2.7	9
		high	54.9	3.6	9
	retracting	mean	37.8	2.5	18
		low	36.1	1.6	9
		high	39.6	2.0	9
	hysteresis		16.6		
2	extending	mean	59.5	5.5	30
		low	56.7	4.5	15
		high	62.3	5.1	15
	retracting	mean	29.8	3.4	30
		low	27.5	2.8	15
		high	32.1	2.1	15
	hysteresis		29.7		

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AN ADJUSTABLE WRITING DEVICE

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There are numerous commercially available writing devices, as well as specially adapted writing devices fabricated by the occupational therapist. The variety of devices reflects their importance to clients lacking strength and fine motor skills required for writing or other office skills. The Departments of Rehabilitation Engineering and Occupational Therapy at the Rehabilitation Institute of Chicago have designed and developed an adjustable writing device for use by a client with a spinal cord injury.

In reviewing commercially available or specially adapted writing devices, the main disadvantages are: 1) the user may require the assistance of another individual to change the various writing tips, 2) on many of the writing devices the pen holder has a fixed angle. The new adjustable writing device promotes independence by allowing insertion in a holder of a definitive wrist hand orthosis (Fig. 1). Additionally the device can be angled and adjusted with an allen wrench. The set-up for this device is similar to the modular mouthstick system which was developed by the Northwestern University Rehabilitation Engineering Program and Rehabilitation Institute of Chicago Occupational Therapy department. The components needed for this device are commercially available as a kit. The client is then free to independently change a variety of tips, such as pen, pencil, or eraser.

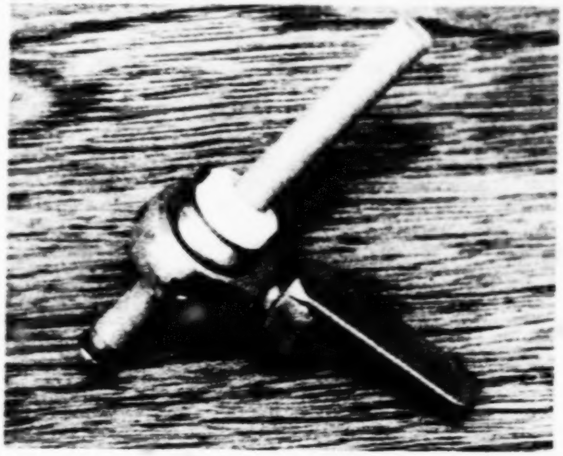

Fig. 1. Adjustable Writing Device

Fabrication of the adjustable writing device can be performed with standard tools found in most shops. The main component of the design is a control rod end. Two 8-32 set screws are installed so that the angle of the raceway can be fixed. After shortening and boring of the control rod end's threaded shaft to decrease the weight of the device, it is inserted into an 8-32 hex nut which is brazed to a standard orthosis tab. Loctite is used to attach the control rod end parallel to the orthosis tab. The design is now ready to utilize components from the modular mouthstick kit to provide interchangeability. One of the kit's four appliance tips is cut in half and inserted into the raceway of the control rod end from above. One of the fiberglass arrowshafts provided is cut approximately 3" from the end containing the round head screw and o-ring. This 3" piece is inserted in the assembly such that 1" protrudes below the control rod end.

The three remaining appliance tips can be set up with the appropriate instruments (e.g., pencil, pen, eraser, paint brush). Using the mouthstick kit's stand, the system is now ready to use at the client's work station (Fig. 2).

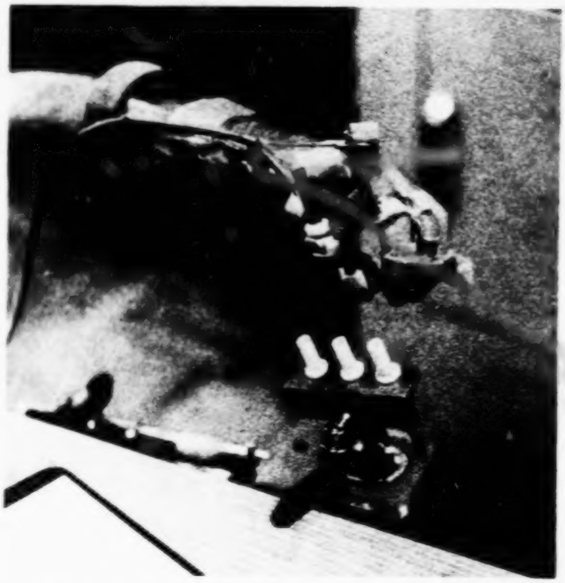

Fig. 2. Completed kit stand mounted on work station

SUMMARY

To promote independence, an adjustable writing device has been designed for a client to use in a vocational, avocational, or school setting. The therapist can evaluate and set the desired angle for specific writing tasks (Fig. 3). In addition, the angle can easily be changed for different activities. The client can choose a variety of tips for the set-up. This device may also be used with a school age child for crayon/coloring activities. The set-up may have potential for use with other diagnoses, such as muscular dystrophy and arthrogyroposis, or as a temporary device for clients awaiting a definitive prosthesis.

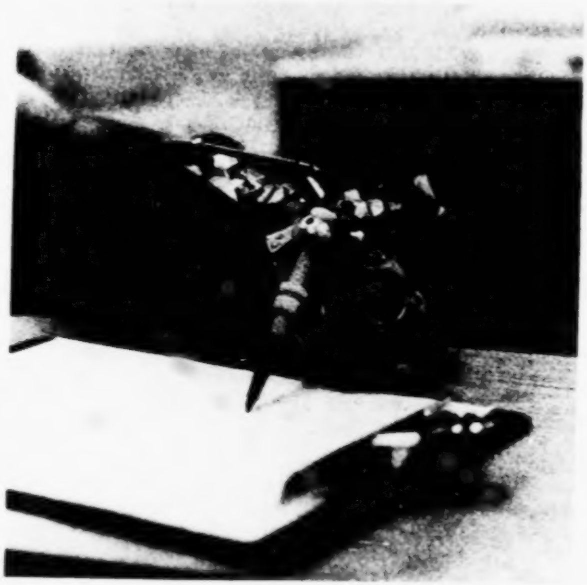

Fig. 3. Completed components of the Adjustable Writing Device

REFERENCE

Kozole, K., Gordon, R., and Mycoff-Schultz, J. "A Modular Mouthstick System." Proceedings of the Fifth Annual Conference on Rehabilitation Engineering, 1982.

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14.5

CAD/CAM SYSTEM APPLIED TO THE FOOT SHAPE FOR THE PROSTHETIC DEVICE

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Yukio SAITO (Tokyo Denki University)

The present paper deals with an automatic measuring method for a human foot shape with a three dimensional coordinate measuring machine which has 5 degrees of freedom. The object of this study is to design a fitted prosthetic device for a disabled person's foot or a good shoe for sports and manufacture these positive models.

To achieve these purposes, it is necessary to measure a foot shape exactly and fully and to operate the measured shape so as to be satisfied various demands. In this paper mainly next three items will be explained.

(1) INTELLIGENT SHAME	to measure a foot shape with a three dimensional coordinate measuring machine.
(2) CONNECTING INDEX	to connect the measured parts smoothly.
(3) BENDING CURVE	to correct the measured and connected shape.

MEASURING MACHINE SYSTEM

To measure a soft object like a foot, an optical measuring method is to be hoped, but to measure exactly and fully, a three dimensional coordinate measuring machne with a touch probe brings a good result. For this reason, a micro-computer based measuring system which can be measured any complicated objects,operated the measured shape numerically and manufactured that positive model. Therefore, this system can be called a CAD/CAM SYSTEM FOR THE LIVING BODY.

Figure-1 shows this system. This measuring machine is a bridge type three dimensional drive machine with two dimensional rotation structures on its Z axis. To measure a foot, this machne has 5 axes (three degrees of freedom X,Y,Z and two degrees of freedomP,Q). And a capacitance type touch sensor with a needle probe is used to obtain an ON/OFF touch signal. The slave computer is a 5 degrees numerical controller with 8 DDA integrators to control its 5 motors. The personal computer is used as a host computer which executes control comands for the slave computer and operates the measured shape.

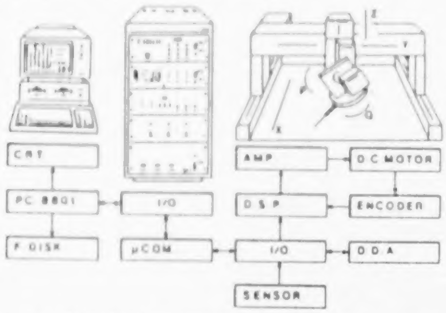

Fig.-1 CAD/CAM SYSTEM FOR THE LIVING BODY

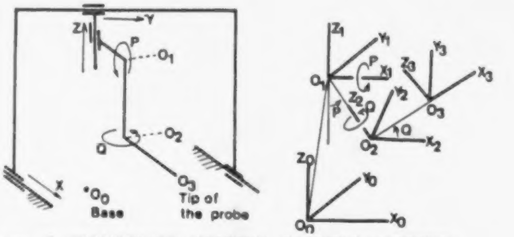

Fig.-2 COORDINATE SYSTEMS OF THIS MACHINE

This coordinate systems can be shown in Figure-2. These coordinate systems may be expressed as follows. O_0-$X_0Y_0Z_0$(base), O_1-$X_1Y_1Z_1$(rotation center P), O_2-$X_2Y_2Z_2$(rotation center Q), O_3-$X_3Y_3Z_3$ (tip of the probe). An obtained value from measuring is coordinate value $\mathbf{X}_1(x_1,y_1,z_1)$ which shows a point O_1 in system O_0 and angles of rotation P and Q. It is necessary to calculate a value $\mathbf{X}_3(x_3,y_3,z_3)$ which shows a point O_3 in system O_0. Where each coordinate transformation matrices are $\mathbf{R}p$(rotation P), $\mathbf{R}q$(rotation Q), $\mathbf{T}r$(axial distance between O_1 and O_2) and $\mathbf{T}s$(lengh of the probe). The coordinate transformation can be expressed as

$$\mathbf{X}_3=\mathbf{R}p\cdot\mathbf{T}r\cdot\mathbf{R}q\cdot\mathbf{T}s\cdot\mathbf{X}_1 \quad \ldots\ldots\ldots(1)$$

And This inverse transformation is

$$\mathbf{X}_1=\mathbf{T}s^{-1}\cdot\mathbf{R}q^{-1}\cdot\mathbf{T}r^{-1}\cdot\mathbf{R}p^{-1}\cdot\mathbf{X}_3 \quad \ldots\ldots\ldots(2)$$

These transformations are calculated in host computer.

MEASUREMENT FOR FOOT

Figure-3 shows an example of the foot shape measurement. To measure a foot, it is necessary to move the direction of its sensor probe, because the slope of measuring cross-section must be changed according to the measuring parts. Therefore, the path and direction for the probe must be setted for the foot shape. Generally, it is impossible to measure automatically, but a new INTELLIGENT SHAME was developped. In this study, at first, a foot plaster model is measured fully and next with its data the probe path model data which is the shortest path and touchs perpendicularly for the shape is generated automatically. The operator only inputs the rough size of measured foot, for example, only shoe size.

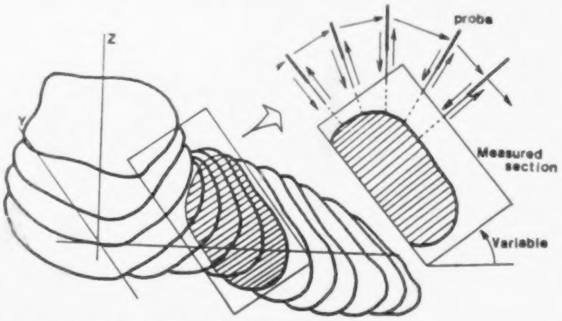

Fig.-3 FOOT SHAPE MEASUREMENT

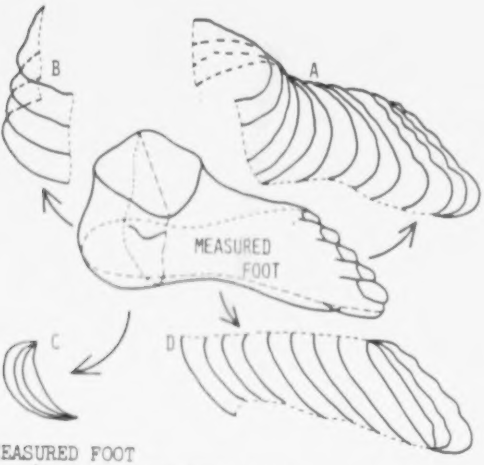

Fig.-4 MEASURED FOOT

Figure-4 shows a measured foot shape. A total shape for the foot can not be measured at the same time. The shape is measured dividing 4 parts, upper front, upper back, lower front and lower back. The lower parts are measured a prastics model after modeling its measured foot.

DATA OPERATION

One of the merit of this system is to be expressed a measured shape numerically. Many operating methods (connection, addition, reduction magnification, interpolation, etc.) were developped under this system. In these methods, important methods are connection for the measured 4 parts and partial correction for the shape.

It is very difficult to connect the 4 parts which have not a standard surface or a point. We gave the attention to the smoothness of the living body shape and developped an original CONNECTING METHOD. Figure 5 shows this method. Any corresponding two measured curves have big errors in the edges. To connect these curves, a strain energy for its curve is calculated. The curve is expressed with B-spline curve segment Ci(t). The partial energy Ei is

$$\mathbf{E}i=\int_{t=0}^{t=1} \frac{\ddot{C}i(t)^2}{1+\dot{C}i(t)^{5/2}}\,dt=\int_{t=0}^{t=1} \ddot{C}i(t)^2\,dt \cdots\cdots(3)$$

And its total energy $\mathbf{E}$ is

$$\mathbf{E} = \Sigma\, \mathbf{E}i \qquad \cdots\cdots\cdots(4)$$

As shown in Figure-6, a plaster foot model was measured and its any measured curve expressed one cross-section was divided into two curves, front and back. And to compare the energy, one curve was moved. Figure-6 shows a that strain energy Ex(X element) and Ey(Y element). Interestingly, if the moving quantities dx and dy also equal to zero, that total energy is the smallest. This result holds true of any parts of the living body. We call this energy CONNECTING INDEX. In actual measurement, the measured curves are connected with this method to minimize this index.

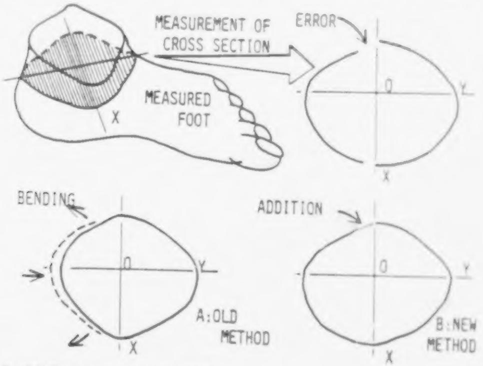

Figure-5 CONNECTING METHOD

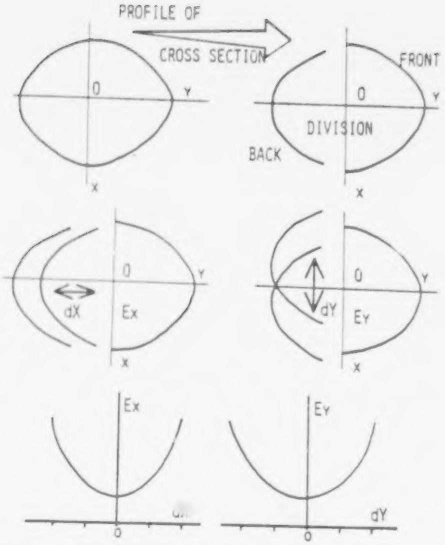

Figure-6 CONNECTING INDEX

Next, it is very important operation for these measured shape to correct the shape partially. This purpose is to change a model for the prosthetic device or shoe model. In this, we use BENDING CURVE EQUATION. As shown in Figure-7, if the measured and connected curve is pulled along the x axis, that transformed quantity U is

$$U= \frac{2\pi dx}{4-\pi}\left(\frac{1}{\pi}-\frac{1}{4}\phi\sin\phi-\frac{1}{\phi}\cos\phi\right) \qquad \cdots\cdots\cdots(5)$$

With this equation, various corrections can be done freely.

These CONNECTING INDEX and BENDING CURVE are fundamental methods to operate the measured foot with these measuring method.

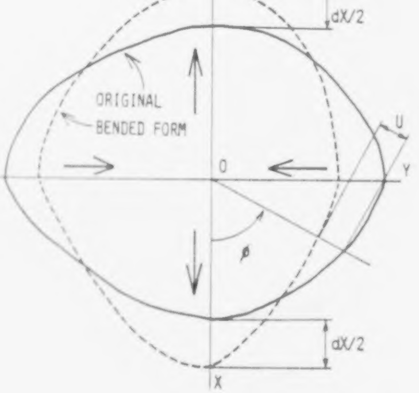

Figure-7 BENDING CURVE

RESULTS

Figure-8 is an one example of measurement and connection. The measured parts are 14 sections in the upper front, 4 sections in the upper back, 13 sections in the lower front and 5 sections in the lower back. And the measuring pitch is about from 5 to 20mm.

Figure-9(a) is one example of measurement, connection and interpolation for the natural foot and (b) is an corrected results for this foot.

Figure-10(a) is a natural foot and (b) is his affected foot.

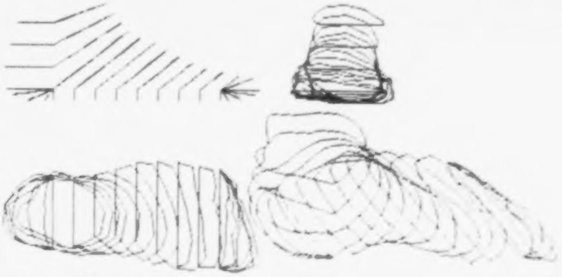

Figure-8 EXAMPLE 1

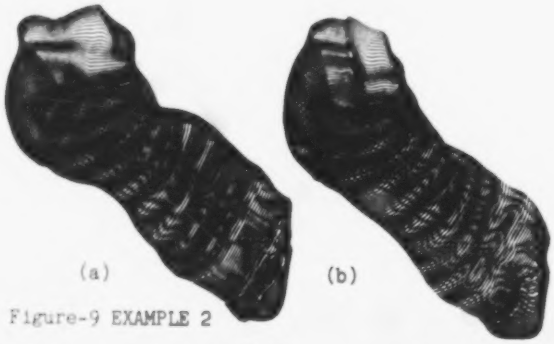
(a) (b)

Figure-9 EXAMPLE 2

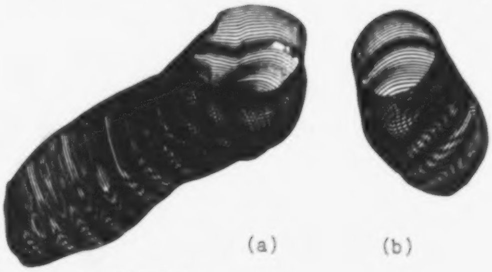
(a) (b)

Figure-10 EXAMPLE 3

CONCLUSION

To make a fitted prosthetic device or a good shoe model, we thought that the most important thing is to measure the shape exactly and fully. And it is necessary to provide the useful operating method. Now, these fundamental methods have been developped. Next we will manufacture a good model with this system.

This system has produced several merits. A complicated and soft object like a foot can be measured automatically,exactly and fully. These system is quite useful in shortning the design time, because designs are fine-tuned numerically, they can be optimized without constructing test model.

This study was performed with the support of the Ministry of Health and Welfare for welfare equipment in Japan 1983.

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PREDICTING THE LOADED SHAPE OF AN AMPUTEE'S RESIDUAL LIMB

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ABSTRACT

CAD/CAM design techniques have made significant technological advances in commercial manufacturing processes possible by allowing the user to consider the interrelation of the many factors which contribute to the mechanical behavior of geometrically complicated structures. Research in the application of these same technologies to the field of prosthetics is beginning to show gains in the reduction of hand labor, possible increases in the average overall quality of fit, and lower overall per unit cost to the amputee. This paper describes work being done at the VA Medical Center and Baylor College of Medicine in Houston, Texas to demonstrate the efficacy of using techniques in the fabrication of above knee prosthetic appliances.

INTRODUCTION

The advent of greater team work between the surgeon, engineer and prosthetist, is a major step toward solving many problems encountered during the rehabilitation of an amputee. During the past two decades many advancements have been made ranging from the development of more functional terminal devices for upper extremity prostheses to more natural control mechanisms (EMG controlled electronic drives) to the use of polymeric materials that permit lighter, more cosmetic prostheses to be fabricated. Biomechanical studies to define socket shapes that account for areas on the residual limb which can support high pressures while reducing the loads experienced on other areas with greater sensitivity and completely protecting regions that are particularly sensitive to mechanical deformations have contributed to more successful fittings and have formed a knowledge base that is used in the prosthetics and orthotics schools in the United States. However, the use of computer technology has not been highly exploited in the socket fitting process, and there is substantial art that may be transferred into quantifiable information that can further enhance the impact of the prosthetics industry on the quality of life enjoyed by the veteran amputee.

Numerical control (NC) machines are now used routinely to manufacture components for prostheses but computer technology has had little impact on the process of fitting a socket to an amputee's residual limb.

The amputee's residual limb is a dynamic structure that assumes different shapes as the amputee performs various activities and these shape changes cause the loading in the artificial limb socket or orthopedic shoe to change. Predicting the shape of the loaded limb so that the residual limb will experience a stress pattern that relieves areas that are sensitive to loading during activity is the primary task of the prosthetist in fitting a socket. In current practice, the socket shaping is the product of the prosthetist's experience and remains an art which requires considerable investments of time. Consequently, this project is focused on studying the art of shaping a prosthetic socket and transforming that art into a more quantitative science through the use of advanced stress analysis techniques, material property measuring systems and biostereometric shape characterizing technology.

BACKGROUND

The current status of the socket fitting is most clearly illustrated in Orthopaedic Appliances Atlas(1) which is used as a text at the Prosthetics and Orthotics School at UCLA. The last major advances in lower extrmity prosthetics appliance fitting technology were the PTB below knee socket and the quadrilateral design for above knee sockets. While these techniques are well founded on basic anatomical and physiological considerations, there are wide variations in the effectiveness of the actual fitting of a particular patient. Successful use of these techniques requires considerable clinical judgements on the part of the prosthetist, particularly when a client is obese or has an excessively scarred residual limb stump.

This work is based on the assumption that the load-deformation behavior of the soft tissue in a limb can be described and analyzed using mechanical analysis techniques that have been developed for the aerospace program and mechanical design markets and are now commonly used tools in engineering practice.

In solving complex material systems arranged in irregular geometries, e.g. the soft tissue on the residual limb, the finite element technique has proven to be a valuable tool that enables designers to predict the mechanical response of the material (its change in shape) to changes in the forces that act on its surfaces. The technique is based on defining a structural network assembly of the physical system so that the system's form can be translated into a mathematical model that is analyzed using a digital computer. Many finite element computer codes are currently available and in this work a system known as ANSYS is used.

There has been a vast interest in developing methods to characterize biological shapes. Many techniques have been developed and are currently in clinical use. Moire fringe techniques are used in early screening for scoliosis(2) and have been advocated for use in characterizing the shape of residual limbs(3). Biostereometric techniques have been developed that provide very precise shape information about body parts or segments(4-5). In fact, these technologies have been applied to characterizing the shape of a residual limb and have been used to measure incremental volume changes that could contribute to the successful fitting of a socket(6-7).

In the areas of soft tissue characterization there is again a vast quantity of literature relating to techniques for measuring the mechanical properties using noninvasive methods(8-11). Many techniques could be cited but the technique that has been developed to satisfy the requirements in this study is the use of ultrasound.

In the area of CAD/CAM studies, two studies have been of particular relevance to this work. The first was done by Research Triangle Institute; in this study the feasibility of using CAD/CAM to produce consumer footwear was demonstrated. At the current time CAD/CAM systems are operational which enable style changes to be transferred into 2-D geometry patterns for cutting shoe uppers and bottom mold design. The second study has been done by Foort *et al* and has demonstrated the feasibility of using CAD/CAM technology to modify standard below knee socket profiles using interactive graphics and a numerically controlled router to create a socket mold. Foort's study is particularly relevant to the proposed work since it represents a demonstration of how prosthetics can use computer technology to reduce space requirements for storing information that is helpful when a new socket is needed for an established prosthesis user. It also demonstrates that satisfactory sockets can be fabricated using NC technology. The proposed work in this study will make it possible to reduce the skill level that must be exercised by the prosthetist in interacting with the computer.

MATERIALS AND METHODS

Ultrasonics are used to define the mechanical characteristics of the materials comprising the anatomical segment and to define the internal structure, i.e. location of the skeleton within the soft tissue mass. Specifically, an ultrasonic doppler is utilized to provide these necessary data. The technique involves using ultrasound to measure internal displacements of the tissue that result from the application of an external, cyclic mechanical perturbation of the tissue. In this application of ultrasound, the basis equation relating the frequency shift of the doppler signal, Δf, to the velocity of the scattering particles, $\bar{v}$ is

$$\Delta f = \frac{2f_o}{c}\,\bar{v}\;.\;\bar{d}, \qquad \underline{1}$$

where f_o is the center frequency of the transmitted ultrasonic wave, c is the speed of compressional wave propagation at frequency f, and $\bar{d}$ is a unit vector pointing in the direction of the ultrasound beam (Baker, Forster and Daigle, 1978)(12).

The device used in this work makes use of a vibrating head which has several ultrasonic transducers embedded in it. These transducers are used to triangulate tissue particle velocities at various locations. A minimum of 12 transducers making 27 measurements would be required to fully characterize the viscoelastic motion of any given depth, but by restricting the motion of the head to combinations of pure compression and pure longitudinal shear, and by assuming that the tissues can be idealized as isotropic, this formulation can be reduced to 3 transducers making 8 measurements.

The current model of the contourgraph is manually operated but was designed so that the operation can be automated and simplified. Currently it produces a graphical record to the different contours, and a printout of the numerical values for each cross-sectional area, incremental length, and the total stump or socket volume. This unit offers the following characteristics:

a. The design of the sensor allows it to also be used as a comparator.
b. The cup upon which the residual limb rests is free floating relative to the base of the sensor. This enables changing the paper upon which the contour tracings have been made without disturbing the residual limb. This is important when multiple scannings are to be made since it is unnecessary to realign the residual limb.
c. The coordinate system is polar.
d. It can be motorized to further simplify data acquisition and alleviate the problem of instability of the "feeler arm" with angular accelerations.
e. The sensor also functions as a planimeter so that when a tracing of a cross-section is completed, there is a direct readout of the area of that cross-section.
f. The diameter of the entire sensor is only 33cm. This makes taking measurements more comfortable for the standing patient. The wider based stereometric form sensor required the patients to support themselves in positions which were difficult to maintain during the contour tracings unless they were seated rather than standing.

DISCUSSION

It is envisioned that the practitioner will use coloring, such as finger paints, to mark sensitive areas in a manner similar to how prosthetists currently mark bony prominences as land marks on a case. An optical sensor mounted on the shape sensor transfers this boundary value information to the grid as it is generated for use in the computer analysis of the desired deformed shape for the segment.

Currently the data that define the surface and the load sensitive areas on the body segment are used to customize pregenerated grids for above knee amputations. By using these techniques, the need for interactive graphics can be minimized and it is possible to avoid this mode of data input entirely in the algorighm that is made available clinically. The interactive mode would only be used to guide the prosthetist through the data collection and to query for additional information when apparent faulty data is entered.

To evaluate the feasibility of using this type of methodology to predict the deformed shape of a residual limb, an analysis has been conducted. In this analysis a stump of 9" length was considered. The unloaded geometry of the stump was estimated from the photographs of the cross sections of the human cadaver leg taken at the University of Kansas. At the University of Kansas, a 95 percentile human cadaver was put in a block of ice; slices of 2cm thickness were cut, and were photographed with a stereometric camera. These photographs, enlarged to full scale, were used in a previous feasibility analysis, to obtain the quantified data of the unloaded stump.

The region of interest was divided into a number of finite three dimensional elastic solid elements. These 3-d solid elements are a family of multifaceted elements consisting of a basic tetrahedron geometry.

The fineness of the grid size increases the accuracy of the results, but the required core size and run costs also increase simultaneously. An iterative grid correction is performed to achieve the desired accuracy, and there are no zero volume elements which would induce singularities. In addition, elements with high aspect ratios are avoided. The entire stump volume is divided into 287 elements.

The model allows the specification of forces and displacements at the nodes and pressures on the element faces. In the initial analysis, the bone was fixed at the top in all directions where as the soft tissue at the top was constrained in only the lonitudinal (z) direction. For the purpose of initial simulation a node located on the outer surface at the midlength of the stump was displaced in the x and y directions to simulate a 0.57" deep radial displacement of the stump.

RESULTS

In order to compare the results of the model with that of an amputation stump, contour tracings of the stump were taken at different levels using contourgraphs. The model was simulated using the ANSYS program with the boundary conditions discussed above, for a static analysis. The simulation results are in good agreement with the experimental results obtained using the contourgraph (variations of less than 1/16") demonstrating the feasibility of using the finite element model to predict loaded shapes of anatomical segments.

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15.1

ELECTROCUTANEOUS SENSORY AUGMENTATION AFFORDS MORE PRECISE SHOULDER POSITION COMMAND GENERATION FOR CONTROL OF FNS ORTHOSES

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ABSTRACT

An electrocutaneous display consisting of five equally spaced electrodes affixed to the skin of the upper arm was used to provide supplementary sensory feedback of shoulder position. Performance scores for tracking tasks in which subjects were required to adjust the position of their shoulder in accordance with a moving CRT cursor, were significantly better when the electrocutaneous supplemental feedback was provided as compared to when it was withheld. These results suggest that the provision of supplemental shoulder position information to users of FNS orthoses systems that are based on shoulder position control may enable them to produce more precise control signals and therefore achieve improved orthosis function.

INTRODUCTION

The transduction of voluntary shoulder movements has been demonstrated to be an effective means by which individuals with C5 and C6 functional level spinal injury can control FNS (Functional Neuromuscular Stimulation) orthoses to provide hand prehension and release. A ball and socket device that attaches to the sternum with double faced tape, and has a telescoping wand that similarly attaches to the shoulder provides two axis of command information. The usual implementation for this tranducer has been to utilize horizontal motion along the axis of protraction-retraction to provide a proportional signal that changes the hand from fully opened to fully closed. Rapid vertical shoulder movements are sensed and converted to logical control signals that allow the user to disengage and re-engage the horizontal axis proportional control (1).

The total protraction-retraction excursion that the user can produce is accessed during initial laboratory set up procedures. An operational range (arbitrarily defined as equal to 50% of this maximum excursion) is specified and the system gain is adjusted as necessary. This procedure eliminates the need for the user to constantly produce very large changes in shoulder position which could be fatiguing and interfere with overall postural stability while the user is seated in a wheelchair. Additional flexibility afforded by the system permits the user at any time to respecify the shoulder position (offset) that corresponds to this zero or starting point of the command range. Frequent reselection of the "zero point" can degrade the user's ability to control his orthosis, however, because he foregoes the benefits of past performance and the familiarity that would apply if he consistently utilized the same range. Having selected a zero point, the user must then rely on his natural joint position kinesthetic sense to aportion the range of motion into equal increments. We set out to study how well a subject could do this and then to determine if the subject's performance could be improved by using supplementary shoulder position feedback from an electrocutaneous display.

MATERIALS AND METHODS

Electrocutaneous display and stimulation parameters

We have determined for individuals with quadriplegia who retain at least C5 or C6 level of function that the skin region along the upper arm or the upper back is satisfactory for the placement of stimulation electrodes intended to produce coded sensations on the skin.

Past studies in our laboratory have shown that with reasonable training subjects can learn to discriminate five different loci in these skin regions from a linear array of five electrodes that are space 40mm apart from each other. This particular array size and electrode separation was selected somewhat arbitrarily but was guided by several observations as follows: When the electrodes in the array were separated by distances closer to the minimum two-point difference limen for successively applied electrical stimulation (cf. 20mm), subjects could distinguish relative changes when the stimulation was switched from one electrode site to another, but they had considerable difficulty in identifying on an absolute basis which particular electrode was active at any given instant. In addition, studies of subjects' average response time for deciding the absolute locus of the stimulation, revealed substantially longer latencies when the total number of electrodes in the array was increased from 4 elements to 8. Preliminary results from those studies suggest that the rate of information transfer is optimum when the electrode array size is limited to 5 or 6 elements. And finally, five elements separated by 40mm can be comfortably accommodated in the space available between the spinal midline and the start of the medial boarder of the shoulder or along the upper arm.

Because it was more convenient to do so, the electrodes for this study were located on the arm. Each electrode was used monopolarly and consisted of a 20mm plastic disk having a round, 3mm diameter stainless steel central contact. A conductive electrode gel was used to provide a stable interface with the skin and an annular ring of double faced tape secured the electrode in place. Great care was exercised to ensure precise replacement of the electrodes on the same skin locus each day. A large conductive pad (41mm X 88mm, "Myocare" - 3M Co.) placed over the deltoid region served as a remote common anodal electrode. Only one electrode from the array was activated at any given time. The subjects were pretrained to be able to identify any randomly activated electrode out of the set with 90% reliability. A 15 Hz signal consisting of bursts of four 50 usec. capacitively coupled constant current pulses (intraburst carrier 1 KHz) was used to drive the electrode display. The intensities of the five electrode channels were preadjusted at the start of each test day to provide equal subjective intensities.

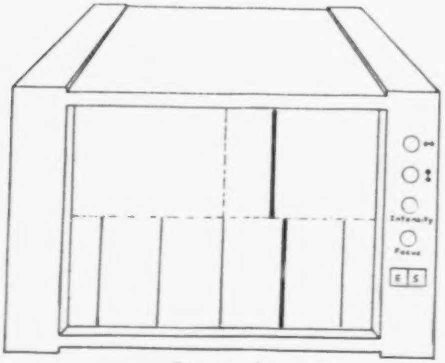


Figure 1

Tracking task and scoring procedure
Two different tracking tasks were utilized. These are referred to as Experiment 1 and Experiment 2 respectively. Both of the experiments utilized a split screen CRT display as diagramed in Figure 1. The signal to be tracked was presented on the top half of the screen while the bottom half was used to present a visual feedback signal when this was desired. The major difference between Exp 1 and Exp 2 was the particular motion that the tracking signal underwent. As will be explained below, Exp 2 consisted of a more dynamic task than was the case for Exp 1.

Experiment 1. For Exp 1 the signal to be tracked consisted of 4 sec static presentations during which a vertical bar at one of five equally spaced target positions was brightened. The particular target position presented on each successive 4 sec trial was selected from a pseudo-randomized list. The target thus appeared to jump from one location to another and the subject had no means of anticipating which target position would be presented next. At the start of each experimental day the subject for Exp 1 (An FNS system user who has C6 level function) was requested to move his shoulder from full retraction to full protraction. This motion was monitored via the shoulder position transducer, and the signal was scaled to represent the distance from the left to the right border of the display screen. For each trial the subject had to move his shoulder to the appropriate position as quickly as he was able. Thirty, 4 sec trials were given consecutively (for scoring the first 5 trials were discarded) and then the subject rested for 2 mins. This procedure was repeated four times and then tracking scores were compiled based on the averages of those 100 trials. Twenty sets of data consisting of 100 trials each were taken for each experimental feedback condition as follows:

1) No supplemental feedback
2) Visual supplemental feedback
3) No supplemental feedback (same as "1" above)
4) Electrocutaneous supplemental feedback

For visual feedback the output of the shoulder position transducer was used to move a bar along the lower half of the video display. This feedback signal displayed the subject's shoulder motion in a continuous fashion. As will be noted again below, it should be pointed out that the visual feedback signal differed in principle from that provided with the tactile display since the latter consisted of only 5 discrete positions. The subject was not always available to work in the laboratory for the same period of time each day, but in general we obtained 10 or 15 sets of data per experiment day.

Two measures of tracking performance were obtained using a modification of a scoring scheme described by Szeto et al.(2). These consisted of ABSOLUTE MEAN ERROR (AME) and TIME ON TARGET (TOT). With the first measure, the distance (expressed as % of full screen) from the target to the tracking signal was sampled at 24 Hz during each 4 sec trial, and an AME was recorded. For the TOT measure, the percentage of time during each trial that the tracking signal was "on target" (i.e. within ±50% of the intertarget separation) was registered. These two performance indices were then further averaged over many tests and the results compared as a function of the various feedback conditions.

Experiment 2. As stated above the tracking task used in Exp 2 was grossly different than that for Exp 1. In Exp 2 the signal to be tracked was moved back and forth across the screen in a continuous manner using a sinusoidal function such that the velocity was maximum at the screen center and diminished to zero at both the right and left borders of the field. The driving frequency used was 1/2 Hz so that the tracking signal required 2 sec to travel from left to right and return. This was defined as one trial. A single test consisted of 30 such trials and lasted 1 min. For scoring purposes the first 5 trials of every test were discarded. A one min rest period was given at the end of each of the 1 min tests. A block of 20 such tests was completed for each of the feedback conditions as described above for Exp 1. Unlike Exp 1, the visual feedback signal that was provided was not continuous but consisted of only 5 possible discrete positions on the screen. Thus, the visual feedback condition of Exp 2 more nearly approximated the situation that pertained for the 5 position electrotactile feedback display.

Another difference between Exp 1 and Exp 2 is that for Exp 2 the range of shoulder motion that was used for the tracking task was limited to 50% of the maximum excursion that the subject could produce, and the "zero point" as explained in the INTRODUCTION was that selected by the subject. This change was made so that the tracking task would cover the same range of shoulder motion that is normally used for the control of the FNS orthoses systems.

RESULTS

Table 1 gives the tracking test scores achieved by the subject for each of the feedback conditions during Exp 1. The first experimental condition "no supplemental feedback" provides control data with which the visual feedback and electrocutaneous feedback may be compared. The third experimental condition when supplementary feedback was again withheld demonstrates what improvement in performance was attained by the subject's having learned to better utilize his shoulder proprioceptive feedback after training with the visual supplemental feedback. Finally, the last case demonstrates the efficacy of using the electrocutaneous position feedback to enhance the subject's tracking performance. While not as good

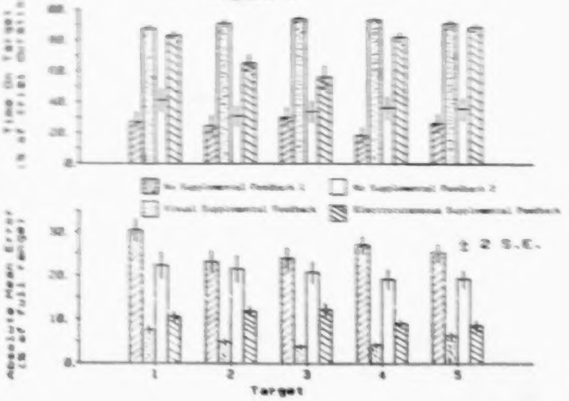
Type of feedback provided	A M E (% test range) ±1 SE	T O T (% trial duration)
No Supplemental Feedback 1	26.1 +/- 0.6	27.5 +/- 1.5
Visual Supplemental Feedback	5.4 +/- 0.2	91.8 +/- 0.4
No Supplemental Feedback 2	20.7 +/- 0.6	35.9 +/- 1.7
Electrocutaneous Supplemental Feedback	10.5 +/- 0.7	75.3 +/- 1.2

Table 1

as the visual feedback in this regard, the electrocutaneous feedback afforded clearly superior performance in comparison to the case where no supplementary feedback was given. Because the visual feedback consisted of a continuous signal rather than a series of discrete steps, however, the visual feedback would be expected to give superior results.

It is possible to extend the level of analysis of the tracking data to determine if the subject was noticeably better or poorer in estimating his shoulder position at any given point over the range tested. This analysis is presented in Figure 2 which shows the average tracking scores attained by the subject as a function of each of the different targets. For the cases when no supplementary feedback was given no gross differences are apparent from inspection of the tracking scores. For the case of the electrocutaneous feedback the TOT data indicate that positions 2 and 3 were more difficult to track. This probably was caused by some fault with the electrotactile display whereby the subject had more difficulty in distinguishing the activation of electrode sites 2 and 3 relative to the other electrode positions. A slight adjustment of the positions of these two electrodes would probably help the subject to attain a more even tracking performance. It should be noted, however, that for the case of every target, the tracking scores when the electrocutaneous supplemental feedback was in use were significantly improved relative to the conditions when "no supplemental feedback" was provided. As shown in Table 1, the subject was "on target" approximately 39% longer with the use of electrocutaneous feedback (p(.001), and his AME score was approximately halved (p(.001).

Figure 2



Type of feedback provided	A M E (% test range) ±1 SE	T O T (% trial duration)
No Supplemental Feedback 1	24.1 +/- 0.3	27.5 +/- 0.6
Visual Supplemental Feedback	11.1 +/- 0.2	65.2 +/- 0.8
Electrocutaneous Supplemental Feedback	10.5 +/- 0.2	65.9 +/- 0.8
No Supplemental Feedback 2	19.9 +/- 0.3	39.4 +/- 0.9

Table 2

The results from Exp 2 in which a sinusoidal signal was tracked also showed that significantly better performance (p(.001) was attained when the subject was provided with electrocutaneous feedback. These data are given in Table 2. Time on Target was improved from 39% to 66% and the Absolute Mean Error was reduced from approximately 20% to 11%. Interestingly, using the visual feedback that was restricted to consist of only 5 positions (Exp 2), the subject's performance was nearly identical to and showed no advantage over the feedback that was provided by the electrotactile display which also contained only 5 positions.

CONCLUSIONS

The results from the two types of tracking tasks demonstrate that tracking performance is significantly enhanced by the use of supplemental shoulder position feedback provided by an electrocutaneous display. To the extent that the utility of FNS systems are dependent on the user's ability to provide precise and consistent control signals by regulating the position of the shoulder, the provision of an electrocutaneous feedback signal should be very beneficial for the user. Future work should include functional tests of prehension skills to support this hypothesis. Then implantable hardware systems should be implemented to enable the feedback systems to be unencombering and reliable. Concurrent studies are presently underway to utilize the same electrocutaneous display described here for the feedback of grasp force information via a frequency moduation code that would be superimposed upon the spatial coding scheme described in the present paper.

ACKNOWLEDGEMENTS

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PERFORMANCE OF A CLOSED-LOOP CONTROLLER FOR ELECTRICALLY-STIMULATED STANDING IN PARALYZED PATIENTS

15.2

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ABSTRACT

A feedback controller for the regulation of electrically-stimulated stance in T4-T11 patients (using intramuscular electrodes) has been evaluated in terms of quality of stance, duration of stand, and ability to compensate for disturbances. This controller consists of four decoupled modified digital PID (Proportional-Integral-Derivative) controllers; one for each knee and each ankle. It has been found that this relatively simple controller effects good standing quality and excellent disturbance resistance, but for an unfortunately short duration (generally 5 to 10 minutes). Analysis of these experimental results points to the need for adaptive features in the controller to obtain longer duration stands.

INTRODUCTION AND BACKGROUND

An initial task in the development of closed loop control of walking is the stabilization of the standing position. Closed loop control should reduce the stimulation of the quadriceps during standing and should allow longer standing. Feedback control of standing should give patients the physical stability and the confidence needed to be functional while standing, without worrying about whether their legs will buckle.

In this paper we describe and critique the performance of a modified digital PID (Proportional-Integral-Derivative) for closed loop control of electrical stimulation during stance. Feedback signals (knee or ankle angles) are obtained by externally-mounted goniometers. Each controller stimulates several muscles in parallel (quadriceps group for knees; soleus for ankles).This controller effectively stabilizes stance with good disturbance resistance for short time intervals, but is not able to obtain (on a repeatable basis) stands of longer than 5 minutes.

Earlier work on a open-loop stimulation controller, a discussion of modified PID controllers for this purpose and a review of related work in the literature appears in (1); a microprocessor implementation of this open loop controller appears in (2).

MATERIALS AND METHODS

Patients in this study had complete lesions from T4 to T11, with no evidence of significant peripheral nerve damage. All were able to use long leg braces for forward locomotion for at least three meters, in a swing-to or swing-thru gait, with the assistance of one person.

Electrode construction, implantation procedure and stimulation methods were as described in (1), except that the intramuscular electrode used 76 um multi-strand stainless steel wire [the number reported in (1) is a typographical error].

Each muscle to be stimulated in FES closed loop standing can be thought of as a single input-single output system. The inputs to the muscle are pulses of varying pulse widths. Since the intervals between pulses are much larger then the pulse widths the input can be approximated as an impulse train. The output of the muscle is force. Because the design of the controller will be digital, it is advantageous to model the muscle in the discrete time domain (3,4).

The control scheme used here for stabilizing standing consists of four parallel controllers. There is one controller for each feedback signal. The angle of the knee or ankle that is being controlled is measured using externally-mounted goniometers (see Figure 1) and compared to a desired setpoint angle for that joint. The controller uses the measured error (that is, setpoint minus measured output) to calculate a corrective input signal which stimulates the muscle, resulting in forces at the knee or ankle. This procedure is repeated during every sampling period. Each of the four controllers effects stimulation of one or more muscles (quadriceps group for the knee controllers and soleus for the ankle controllers).

The actual stimulation delivered to each electrode is limited to a maximum value and, unless no stimulation is directed toward a particular electrode, the pulse-width level is also kept above a minimum value. These minimum and maximum values are specified in the software, and are different for each electrode in each patient. The maximum value is generally chosen for safety reasons, but in some cases it is chosen to avoid undesirable motions (for example, in one patient high levels of quadriceps stimulation results in stimulation of the sartorius as well). The minimum levels of stimulation are specified based upon experimental determination of the pulse-widths required to obtain observable effects.

Since accurate models of the system for control purposes are not available, a modified discrete Proportional-Integral-Derivative (PID) controller was used. A PID controller was chosen because it doesn't make strong assumptions about the system. The PID's proportional term corrects for instantaneous errors, the integral term corrects for accumulated error, and the derivative term detects and corrects for sudden changes in the angles of the knee or ankle. Based on the stability shown in the root locus analysis of the dynamical system during knee hyperextension, the PID controllers is chosen to "lock the knees" (stabilize the knees by keeping them in a slightly hyperextended state). This allows the biomechanical structure of the patient to support most of the standing, reducing the muscle use.

CLOSED LOOP PERFORMANCE

The controller stabilizes the ankles with the help of the patient's hands for support (i.e. holding onto a railing, walker or crutch).

The summation used to approximate the integral term in the PID controller contains a 'forgetting' or 'fading' factor which discounts older data. The use of a fading factor is desirable because it takes into account the fact that some errors might be caused by the patient (i.e. if the patient repositions his body) and should be forgotten. The fading-factor integral term has the form:

$$SUM_k = FF * SUM_{k-1} + ERROR_k$$

where FF is the forgetting factor $[0 < FF \leq 1]$. This fading factor integrator, in the presence of continual error, results in an integral term that does not grow without bound:

$$SUM = 1/(1-FF) * \text{Steady State Error}$$

The error input used by the PID controller is clipped to zero (excluding for negative errors) because of the common occurrance of goniometer slippage. If the measuring devices record negative angles at the knee (or ankle) while the actual knee angle is at a constant zero degrees, the PID's integrator will accumulate a large negative sum . Then, when the knee starts to bend, the controller will be unable to output a corrective pulse width because the integral output (which would be negative) cancels the positive outputs of the derivative and proportional terms (this is called 'integrator wind up'). Another way of approaching this problem is to clip the output of the integrator only.

The PID controller derivative term is used only for positive derivatives. When sudden changes in the knee (or ankle) angle occur, the PID controller outputs a corrective pulse width. Generally, by the time the controller samples the angle again, the knee has returned to the desired setpoint. Thus the PID controller would see a negative derivative which could result in the PID's derivative term cancelling any corrective output of the integrating term that may still be needed. This can produce instability; consequently only positive derivative terms are used.

For both the ankle and knee angles the setpoints of the parallel PID controllers are zero degrees, where extension of the knee is defined as zero degrees. Even though the ankle angle's setpoint is zero, it is assumed that the patient may choose to lean slightly forward. This will result in a steady state error. Since a fading factor integrator is being used, the stimulation of the soleus may reamain below 150 usec (the maximum allowable stimulation) depending upon the proportional term, the integral term, and the fading factor. The resulting stimulation produces some plantar flexion but, more importantly, knee extension.

RESULTS

A typical closed loop trial is shown in figure 2. Note that the patient's knees are "locked", but the ankles have some constant error. This is acceptable since the constant error at the ankles implies the patient is leaning forward.

Figure 3 shows the response of the legs while exerting a perturbing force at the knees. The PID controllers responded as desired with corrective forces when needed. When the knee returned to its "zero" position, the controller didn't abruptly cutoff but instead it slowly reduced the magnitude of the stimulation.

After a relatively short duration, however (generally around 5 to 10 minutes), the PID controller fails to maintain good standing. In most patients this occurs as ever larger levels of stimulation are needed to maintain standing, until the maximum allowable levels for the electrods are reached.

In one leg of one patient, however, after about 10 minutes a "limit cycle" oscillation repeatedly occurs. We believe that this is due to the fact that the minimum pulsewidth needed for motion has increased with fatigue, to a value above the minimum value specified by the controller. Thus when a knee begins to bend, that controller increases stimulation to straighten it but then effectively turns off completely (because it stimulates at a minimum pulsewidth below what is needed to obtain torque). Thus a small disturbance will cause a more abrupt flexion, which the derivative term of the controller will abruptly correct, then turning off.

On the basis ofover 75 experimental trials on five patients, we have reached the following tentative conclustions regarding the ue of nonadaptive PID control for stance stabilization:

1. It appears that PID control alone is not satisfactory for long-duration standing. We have found that abrupt step increases in quadriceps stimulation appears to drive a knee angle error to zero, but that more gradually ramping the stimulation results in the knee not being straightened, even at the maximum pulse-width. This probably arises from velocity-dependent dynamic properties of the muscle. Thus a PID controller at the knee must aggressively respond to even small errors. However, keeping the stimulation at a high level after the knee is straightened leads to rapid fatigue. Therefore it is necessary that the stimulation be "tuned down" (gradually) as soon as the knee is straightened. This is a task that must be carried out adaptively, by a "supervisory" controller such as the one described in (5). The addition of certain adaptive features to the controller greatly extends the duration of stance under closed-loop control, to as much as 75 minutes before maximum levels of stimulation are required to maintain stance; teh development of these adaptive controllers is a topic of current research.

2. In addition, it is frequently the case that the closed-loop control must be initiated from a high level of stimulation (such as the open-loop levels), rather than from minimum values of stimulation. That is, the closed loop controller can start with a standing patient, and

then reduce stimulation as much as possible while maintaining standing; however, if the stimulation begins at a low level, it may not be able to straighten out the knee.

ACKNOWLEDGMENTS

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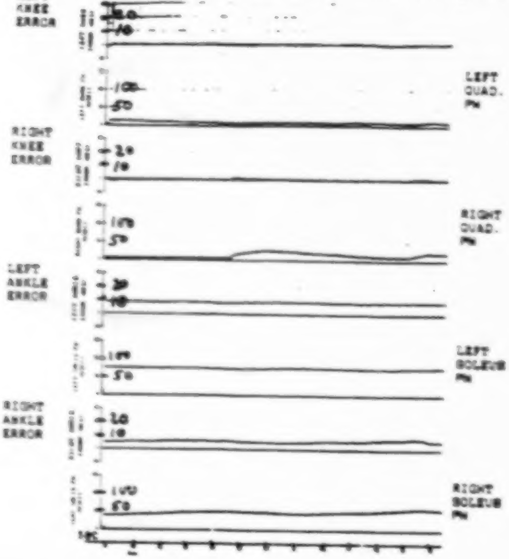
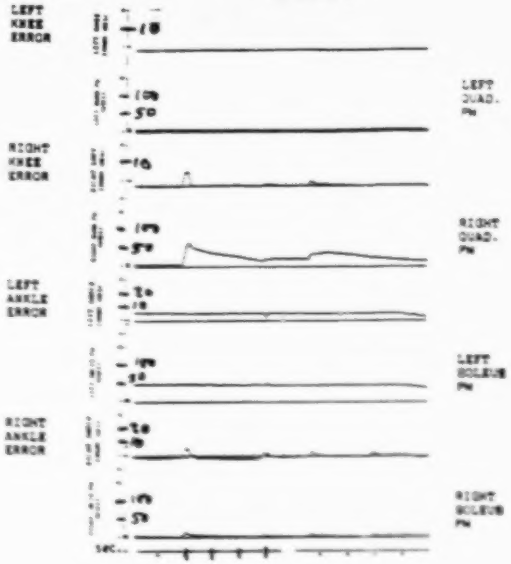


FIG. 2

← FIG. 1

FIG. 3 ↓



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15.3

EFFECT OF MUSCLE STIMULATION ON WEIGHT SHIFTING IN PARAPLEGICS

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ABSTRACT

The effect of stimulating hip and leg muscles, individually and in combination, has been studied for their ability to shift weight. The gluteus medius was found to be most effective in mediolateral shift during walking since it produced no backward movement as did the gracilis, tensor fasciae latae and sartorius. The gluteus maximus and semimembranosus were most effective in bringing the center of pressure forward. Incorporating these muscles and the plantar flexors into a stimulation pattern doubled the stride length and quadrupled the speed of electrically stimulated walking in paraplegics.

INTRODUCTION

A limited reciprocal walking by means of functional electrical stimulation using a walker (1,2) or crutches (3) for support has been achieved in paraplegics with a complete absence of motor function at the level of injury between T4 and T12. The primary problem in these patients has been the difficulty in actively transferring their weight from one foot to the other. Lacking functional hip extensors and abductors and plantar flexors these subjects are transferring their weight with their arms and maneuvering of the upper body. This is a time and energy consuming task which lacks the efficiency provided by the dynamics of normal gait. Using percutaneous intramuscular electrodes (4) we were able to test the effects of stimulation of individual and combinations of muscles of the hip and leg on the movement of the center of pressure.

METHODS

Three subjects with complete absence of motor function and sensation between the levels of T4 and T11 were evaluated. They had percutaneous intramuscular electrodes implanted in hip extensors (gluteus maximus, posterior portion of adductor magnus and semimembranosus); hip and knee flexors (sartorius, tensor fasciae latae and gracilis); knee extensors (vastus lateralis, vastus medialis and vastus intermedius); ankle dorsi flexors (tibialis anterior and peroneus longus); and the ankle plantar flexor (soleus). Two of the hip flexors, sartorius and tensor, also produce abduction of the hip when the knee is in extension (6). Particular emphasis was placed on determining the effect of hip muscles in shifting the center of pressure. Each subject wore long leg braces and stood with one foot on each of two force plates (Advanced Mechanical Technology, Newton, MA, model OR6-5) while holding onto a rail in front of him for support.

First, two subjects (T11 and T8/9) were tested for their ability to voluntarily shift the center of pressure mediolaterally with minimal use of their arms. Second, they were asked to balance themselves before individual muscles and multiple muscles were stimulated to displace their center of pressure. All of the muscles were tested except knee extensors and ankle plantar and dorsi flexors. The subjects were asked to use their arms only minimally; and in the event that they felt that they were falling. Stimulation was a burst of constant current pulses, 20 mA in amplitude, delivered at 25Hz. Data from both force plates on Mx and My moments and Fz, Fx and Fy forces were used to calculate the combined center of pressure. The XY plane was defined to be parallel to the surface of the force plates. A video tape was used to help in the analysis of the movement of their body. Third, tests involving stimulation of multiple muscles were done on a T4 paraplegic to provide active shifting of weight during walking. Two stimulation patterns were programmed into a 32-channel portable microprocessor controlled stimulator (5). The first pattern included knee extensors, hip flexors and dorsi flexors only. The second pattern was the same as the first, but also included hip extensors and abductors and plantar flexors. The subject was timed walking a 15m course and the stride length was measured.

RESULTS

All of the muscles tested produced some transfer of weight. In the mediolateral direction transfer of weight was achieved faster by muscle stimulation than by voluntary effort. Generally the adductors produced ipsilateral movement of the center of pressure, abductors produced contralateral motion, flexors shifted the center of pressure backward and extensors forward. The gluteus medius, tensor fasciae latae and gracilis were the most effective in shifting the weight laterally. Forward shift of the center of pressure was most effectively produced by the semimembranosus and gluteus maximus. Greater shifts were produced when muscles were used in combination.

Figure 1 shows the mediolateral shift in the center of pressure produced by individual and a combination of muscles. The left gluteus medius produced a shift of about 8 cm to the right in 1 second. The center of pressure during stimulation of the gracilis and posterior adductor always shifted first to the contralateral side, as seen in the figure for the gracilis, while the final shift was ipsilateral. This double shift of the center of pressure sometimes occured with stimulation of abductors. However this phenomenon was not observed in the body movement as seen in the video tape.

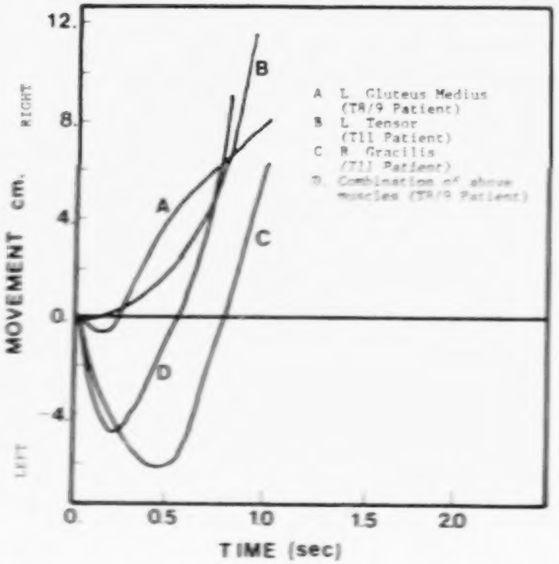

Figure 1. Mediolateral Shift in Center of Pressure Produced by Individual and a Combination of Muscles

Figure 2 shows the effect of individual and a combination of muscles on the sagittal plane movement of the center of pressure. Hip flexors including tensor, gracilis and sartorius (not shown) produced a backward shift in the center of pressure. The gluteus medius, a primary abductor, produced a forward shift in the center of pressure. Muscle stimulation produced a faster shift in the center of pressure than voluntary effort.

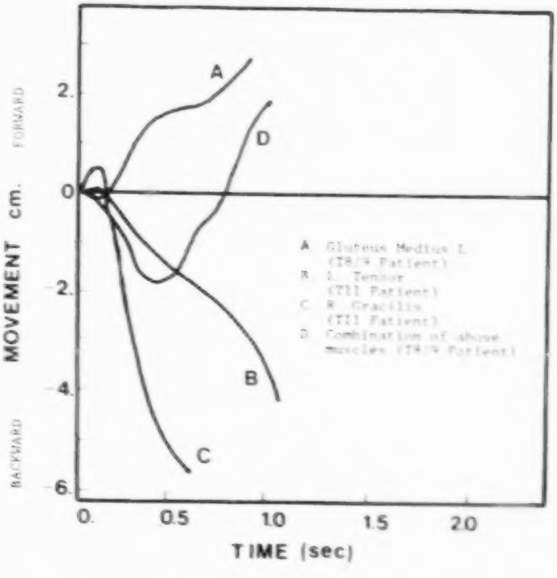

Figure 2. Sagittal Plane Movement of Center of Pressure Produced by Individual and a Combination of Muscles

The video tape observations showed that in most cases upper body movement caused by muscle stimulation was large enough that the subjects would have fallen had they not prevented it with their arms. In the mediolateral direction this corresponded to the point when the upper body was well above the supporting leg.

In the walking experiment, inclusion of hip extensors and abductors and plantar flexors provided subjects with larger stride lengths and increased speed. The stride length increased from an average of 64 cm (s.d. = 18 cm) without these muscles to an average of 1.4 m (s.d.= 10 cm) with the muscles. The average speed increased from .1 m/sec to 0.4 m/sec.

DISCUSSION

Much greater shifts in the upper body were observed in the video tapes than were indicated by the center of pressure measurements. In addition, in some cases, the center of pressure initially moved in the opposite direction to the observed movement of the body. Both of these observations point out the limitations of using the center of pressure measurement as an indication of the location of the resultant force of gravity. In dynamic situations, such as when the subject first begins to move, and in situations when the subject is supporting himself externally with his arms, the center of pressure is not an indication of the center of gravity location. However, when used in conjunction with video tapes, it is useful to compare the time and amount of center of pressure shift with and without muscle stimulation. Despite its limitations, the center of pressure measurement has been useful in understanding the ability of different muscles to shift body weight.

The speed of reciprocal walking in paraplegics will always be limited by the amount of time it takes to shift weight from one foot to the other between steps. The ability of the subject to shift weight also limits the allowable step length. The results showed that weight shifting can be achieved in a much shorter time by stimulation of the hip extensors, abductors and adductors, and plantar flexors than by voluntary effort. Combinations of hip abductors and adductors were most effective in shifting the weight mediolaterally. However, due to their displacement of the center of pressure backward, as is the case with tensor, gracilis and sartorius, only the gluteus medius was used for mediolateral shift of weight during walking. The measurements of center of pressure and stride length measurements indicated that the combination of hip extensors and the push-off provided by plantar flexors were most effective in the forward shifting of weight.

Combinations of hip abductors and extensors and plantar flexors were very effective in increasing both the stride length and speed of walking. The potential for further increases exists both by optimizing the relative timing of muscle stimulation and by testing and incorporating different muscles.

ACKNOWLEDGEMENT

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STIMULATING THE QUADRICEPS AND HAMSTRINGS WITH SURFACE ELECTRODES

15.4

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ABSTRACT

Motor points of the quadriceps and hamstrings were identified in 10 normal subjects for each muscle group by measuring the isometric twitch response at 40 cells of a superimposed grid. Results were compared at 15, 45 and 75 degrees of knee flexion. Bipolar stimulation at 6 pairs of cells was also performed to determine the effect of electrode size, waveform and polarity on torque production. The quadriceps was found to have a motor point over the femoral nerve and a second broader region of excitability over the rectus and vastus lateralis. Neither of the motor points were significantly affected by knee flexion. Motor points of the hamstrings were more variable and were affected by knee angle. Electrode size had little or no effect on torque with monopolar or bipolar stimulation. Polarity often had a significant effect on torque with monophasic waveforms but had no effect with biphasic waveforms. Greater torque was always produced with a biphasic waveform when compared with a monophasic waveform of either polarity.

INTRODUCTION

Electrical stimulation of the lower extremities is currently used in a number of therapeutic applications. It is used to maintain muscle bulk and prevent osteoporosis in SCI patients (1,2), strengthen atrophied muscle (3,4), prevent atrophy during immobilization (5), and as an adjunct to gait training (6).

Despite the proliferation of clinical applications in recent years, there is very little quantitative data available in the literature upon which clinicians can make decisions regarding placement of electrodes and stimulation parameters. The purpose of this study is to provide quantitative data about motor points and techniques of stimulation for the quadriceps and hamstring muscle groups that will be useful to the clinician. In addition, information about polarity and the use of monophasic versus biphasic stimulation are presented which establish certain general principles that may be applicable to most large muscles.

METHODS

Instrumentation

Isometric torque at the knee was measured by an instrument designed and fabricated in our laboratory. Output from the torque transducer was fed to a Gould 2200S strip chart recorder and an A/D converter from which data was acquired by a Digital Equipment Corporation MINC computer.

Stimulation Parameters

A computer-controlled, microprocessor based stimulator, designed and constructed in our laboratory, was used in these studies. The stimulator generated a constant-current waveform, either a capacitively-coupled monophasic waveform or a symmetrical biphasic rectangular waveform. In each case, the duration of the initial unidirectional phase was 300 microseconds. Except where indicated, the stimulus amplitude was set at 60 milliamperes, a level which has been found to produce about 30% of the maximum voluntary contraction of the quadriceps at a frequency of 35 Hz with the knee flexed to 60 degrees.

In his study, only isometric twitch torque was recorded. Because of the large number of measurements recorded during each session, tetanized contractions would have been too fatiguing. Twitch torque was measured by applying a train of 12 pulses, each separated by two seconds. To allow for subject accommodation, the first two twitches were ignored and the peak values of the last ten twitches were averaged to obtain the peak twitch torque for a given set of stimulation parameters and electrode positions.

Electrodes and Electrode Placement

A grid of 40 cells of equal area (8 rows by 5 columns) was drawn over the muscle group to be tested. For the quadriceps, the grid extended from approximately 6 cm below the anterior-superior iliac spine to the midpoint of the proximal patella. The width of the grid was determined at the midpoint of this line by the medial and lateral borders of the quadriceps. For the hamstrings, grid length extended from the ischial tuberosity to a point in the popliteal crease midway between the tendons of the biceps femoris and the semimembranosus. The width was determined as before at the midpoint of this line and was equal to the distance from the medial to lateral borders of the hamstrings.

The rows of the grid were designated by the letters A through H, running proximal to distal. Columns were identified by number, 1 always indicating the most medial column and 5 indicating the most lateral column. Grid cells were thus identified by a letter/number designation; e.g., cell A1 is the most proximal and medial cell.

Electrodes were always placed so that the center of the electrode was over the center of a particular cell. Two sizes of electrodes were used for stimulating the tested leg: a "small" electrode 4 cm x 5 cm (Staoderm-K Electrode, Staodynamics, Inc., 1225 Florida Ave., Longmont, CO 80501) and a "large" electrode measuring 4 cm x 9 cm (Muscle Stimulation Electrode, 3M Myocare, Medical Products Division, 3M, St. Paul, MN 55144).

Monopolar and bipolar stimulation were used. During bipolar stimulation, electrodes were placed on two cells of the tested leg. During monopolar stimulation, an electrode was placed over one of the cells and a second electrode was placed on the opposite leg.

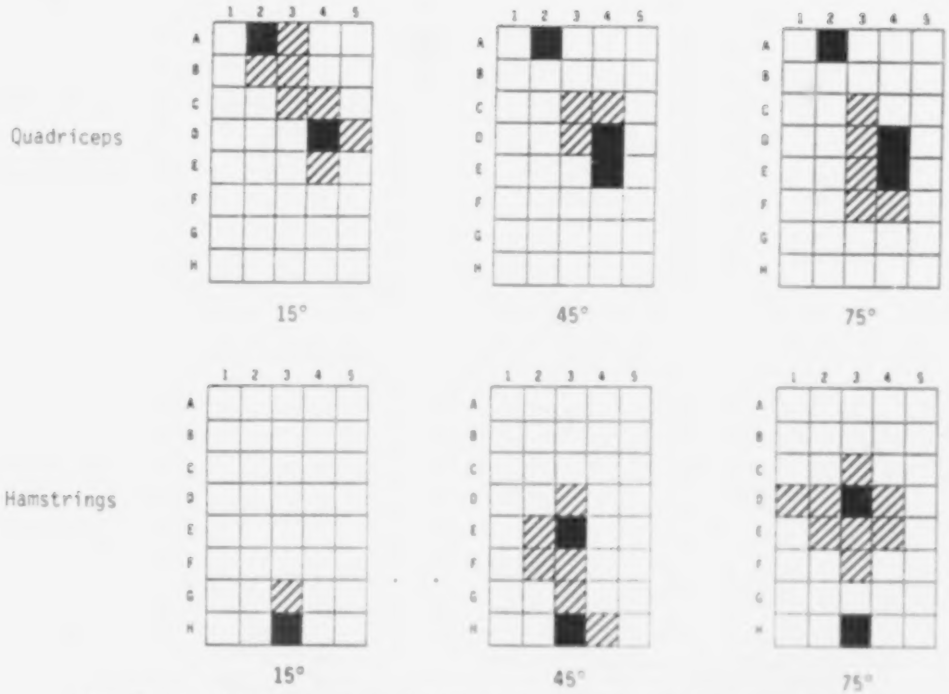

Figure 1. Schematic representation of averaged torque profiles at knee angles of 15°, 45° and 75°. Peaks are designated by the darkened cells. All other cells which have average torques greater than 80% of the torque at the secondary peak are cross-hatched. In the case of the hamstrings at 15° which has only one peak, only the one cell that is within 80% of that peak is cross-hatched.

Subjects

For each muscle, 10 non-obese subjects were used, 5 males and 5 females, ranging in age from 23 to 33 years of age. For the quadriceps study, subjects were seated in a slightly reclined position with the hips in 120 degrees of flexion. The subjects participating in the hamstring study lay prone on a padded bench with the hips in a neutral position. In both studies, the knee of the leg to be tested was maintained in one of three positions: 15, 45 or 75 degrees of flexion.

RESULTS

Monopolar Stimulation (Monophasic Waveform)

The effect of knee angle on torque and motor point location is summarized in Figure 1. Grids are shown for the quadriceps and hamstring muscles at knee angles of 15, 45 and 75 degrees. Peak twitch torque values were averaged over the 10 subjects at each grid cell, and the regions of highest torque are indicated. Black squares indicate a localized peak and cross-hatched squares indicate those cells in which the torque values are within 80% of the peak values.

There are two peaks in the torque data for the quadriceps: a primary peak located over the femoral nerve inferior to the inguinal ligament (cell A2) and a secondary peak over the rectus femoris and vastus lateralis muscles (cells D4 and E4). In absolute values, the average peak twitch torque is 14.9 Nm at A2 and 7.1 Nm over D4/E4. The primary peak is very localized, and it was found that a 2 cm shift in the location of the electrode could cause a profound change in torque output. The secondary peak is less than half of the primary peak in amplitude but covers a much broader area so that positioning of the electrode in this region is far less critical.

From Figure 1, it is seen that the location of the motor points of the quadriceps are not significantly affected by knee angle between 15 and 75 degrees of knee flexion. The primary peak occurs at A2 for all three angles and the secondary peak shifts only slightly from D4 at 15 degrees to D4/E4 at 45 and 75 degrees. Although the secondary peak shifts only slightly, the 80% region around the peak does shift from a proximal to a distal location as the knee is flexed from 15 to 75 degrees.

The absolute torque is, however, very dependent upon knee angle. The average torque at the primary peak (A2) is 10.7, 14.9 and 18.2 Nm at 15, 45 and 75 degrees. At the secondary peak the corresponding torque values are 4.5, 7.1 and 7.4 Nm.

It is also seen from Figure 1 that the hamstrings are most responsive along the midline of the distal portion of the leg. There are two peaks of about equal magnitude at E3 and H3. The average peak twitch torque at the proximal motor point (E3) was 4.1 Nm and at the distal motor point (H3) was 4.2 Nm.

Knee angle has a profound effect on the motor points of the hamstrings. At 15 degrees, there is only one motor point (at H3). This motor point does not shift with knee angle, but a second motor point is seen at 45 and 75 degrees at E3 and D3 respectively. The absolute torque values at the distal motor point are 6.0, 4.2 and 1.9 Nm at 15, 45 and 75 degrees. For the proximal motor point, the corresponding numbers are 4.1 and 1.9 ft-lb at 45 and 75 degrees.

Bipolar Stimulation (Monophasic and Biphasic Waveform)

In the bipolar studies, both electrodes were placed at one of six selected sites on the test leg. They were chosen to provide a variety of electrode placements with regard to orientation and spacing.

The effect of polarity on torque output is summarized in Table I. Shown are the average percentage changes in torque when polarity was reversed for monophasic and biphasic stimulation using both small and large electrodes. Also shown below the average changes are the range of the percentage change in each situation. As one would expect, the percentage change was much larger when using a monophasic waveform than when using a biphasic waveform. Using a monophasic waveform, the average percentage change was greater than 20% but the result is highly variable, ranging from no change in some cases to changes of more than 100% in others.

	Monophasic		Biphasic	
	Small	Large	Small	Large
Quadriceps	29 (0-112)	21 (6-75)	6 (0-20)	4 (0-14)
Hamstrings	24 (0-87)	9 (0-18)	5 (0-18)	7 (0-14)

TABLE I. Effect of polarity change. Means of the absolute values of percentage change in torque when polarity is reversed using monophasic and biphasic waveforms and small and large electrodes. Ranges are shown in parentheses.

With symmetrical biphasic waveforms, torque output is not significantly affected by polarity. The average percentage changes and the ranges shown in Table I are what one would expect given the inherent variability in electrically induced torques under these experimental conditions.

Torque generated by monophasic and biphasic waveforms are compared in Table II. In each case, torque produced by using a biphasic waveform is compared with torque produced with a monophasic waveform, with polarity selected to give the largest monophasic output. It is seen that even after selecting polarity to yield the highest torque output, switching to a biphasic waveform resulted in a significantly higher torque output (on the average, a 20-25% increase). This was true for both muscles, with both sizes of electrodes, and over all electrode locations tested.

The final comparison involved electrode size. With both monopolar and bipolar stimulation, there was no statistically significant difference in torque production with the two sizes of electrodes used in this study.

	Small	Large
Quadriceps	24 ± 13	25 ± 10
Hamstrings	22 ± 18	23 ± 12

TABLE II. Monophasic versus biphasic waveforms. Means and standard deviations of percentage increase in torque for small and large electrodes when using a biphasic waveform compared with a monophasic waveform with polarity selected to yield the highest response.

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15.5

INPUT-OUTPUT RESPONSE OF THE QUADRICEPS MUSCLE IN PARAPLEGIC PATIENTS

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ABSTRACT

This work describes the system input/output response properties of electrically stimulated quadriceps muscles observed in paraplegic patients using intramuscular electrodes. These experiments involved four paraplegic patients, with complete spinal cord lesions from T4-T11. Output muscle response to stimulation by constant current amplitude, constant frequency pulse trains of varying pulse widths was investigated. The relationship between stimulus pulse width and knee joint angle was examined to characterize the dynamic response of the system. It was found that each electrode-muscle-joint system can be well modelled as a second-order dynamic system in cascade with a memoryless nonlinearity,and this was confirmed by comparison of computer simulation of the models with experimental data.

INTRODUCTION

The use of functional neuromuscular (electrical) stimulation (FNS) by means of percutaneous electrodes has proven to be a potentially useful experimental orthosis for both paraplegic and quadriplegic patients (1). Orthotic devices are being developed to restore some functional use of paralyzed muscle.

Essentially, any FNS control system must have stable, repeatable, regulated muscle input-output properties over a wide range of muscle lengths, fatigue, and electrode movement. It is believed that such results are possible using closed loop control (2), which modulates the input stimulus pulse width based upon the error between the desired movement and the actual movement. To generate this required input stimulus, the controller must have knowledge of the input-output parameters of the electrode-muscle-joint system. These parameters can be identified from the knee angle output of the system to various input stimulations.

The (electrode-muscle-joint) system can be adequately modelled, for control purposes, as a second-order discrete time linear system cascaded with a nonlinear memoryless gain (2), as shown in Figure 1. The nonlinearity is the experimentally observed relationship between the stimulus pulse width and the steady-state joint angle produced during a constant amplitude and frequency pulse train. A closed-loop controller must account for this nonlinearity. The dynamic model parameters were determined from an examination of the transient response of the electrode-muscle-joint system. The "rise time","settling time", and steady-state output of the system in response to step inputs was used to calculate the model parameters. The model obtained was confirmed by comparing its predicted response (in computer simulation) with experimental results.

METHODS

A MINC 11/23 laboratory computer (Digital Equipment Company) was used both to control an input stimulator and to collect output angle data from a CYBEX II (Lumex Inc., Bay Shore,N.Y.), a commerically available dynamometer. The quadriceps muscle of the patient was stimulated a single percutaneous electrode using various stimulation patterns (1).

The nonlinear relationship between the steady-state (angle) output of the knee and the pulse width of a constant amplitude (20mA), constant frequency (25Hz) stimulation sequence was determined experimentally for 4 different pulse widths (25,50,100,150). Three different resistances for the quadricpes to work against (30,60 and 90 degrees/second settings on the CYBEX II) were used in determining this nonlinear relationship. The threshold value of the nonlinearity (that is the minimum pulse width value that causes a measurable angle change) was obtained using a ramp stimulation pattern (linearly increasing PW at constant amplitude and frequency), from zero to 150 microseconds over a duration of 6 seconds. The system's transient properties were evaluated from its response to step increases in pulse width. The resistance setting was kept constant at 60 degrees/second for both the ramp and step inputs.

For each of the above stimulation patterns, the observed responses were repeatable to within 9% (in 580 total experimental trials). There were 3 trials for which this percentage was greater than 9% --these trials were excluded from anaylsis. Averages of three identical trials were used in the parameter identifications, the threshold observations, and the determination of the nonlinear relationships described above.

RESULTS AND DISCUSSION

In this work, two electrodes per subject were used in the model identification. The threshold value for each individual electrode-muscle-joint system nonlinearity showed variation among patients as well as between electrodes in a single patient. The value of the threshold was typically between 20 and 50 microseconds, although threshold values as low as 5 microseconds and as high as 60 microseconds were encountered. In Figure 2, a typical experimental plot of the nonlinearity relating the steady-state knee angle to constant pulse width input is shown, for two resistance settings of the dynamometer free-swinging level arm. Note that there are regions of steep slope and flat slope. The shape of the nonlinearity is similar for all four patients, although the amplitude of the angle varies from patient to patient. In the large pulse-width range, gradually decreasing or near-constant slope is evident in most patients. This experimentally-determined nonlinearity is similar in shape to the muscle fiber recruitement gain reported in the literature (3), but may also reflect biomechanical factors.

The knee angle response to a step input stimulation pattern is shown in Figure 3. This curve is characteristic of a second order dynamical system response, with a rise time, a settling time, and an overshoot, as indicated. From measurement of "rise time", "settling time", percent overshoot, and steady-state gain, the parameters of the second-order discrete-time (sampled data) dynamic model were chosen using standard methods (4). The step response of the electrode-muscle-joint systems was,in most cases, underdamped. The duration of these oscillations did not exceed 2 seconds in any individual output response. In some cases, especially with stimulus pulse width values ranging from 100 to 150 microseconds, the system showed no oscillations, which is characteristic of an overdamped system. The settling time was defined as the time interval necessary for the output response to remain in a window of +/- 2% of the steady-state joint angle. In all cases the settling time was less than or equal to 1 second. The rise time, defined here as the time interval from 10% of the peak response to 90% of the peak response, was generally less than 1 second with the maximum rise time for all patients not exceeding 1.0 (± 0.2) seconds at a constant resistance of 60 degrees/second.

Using the rise time, T_R, and the settling time, T_S in response to a step input, the coefficients of the following second order differential equation:

$$\ddot{x} + 2\delta w\dot{x} + w^2 = u(t)$$

were found using the relationships

$$T = 4 / w\delta \qquad T = (0.8 + 2.5\delta)/w$$

in (4; pp.316-7). Here w is defined to be natural frequency and δ is the damping ratio. Simultaneous solution of these equations yields w and δ.

With the CYBEX II resistance held constant at 60 degrees/second, the damping ratio did not exceed 3 and was generally about 1. That is, in all cases the system was underdamped. The value of the natural frequency ranged between 3 and 7, with the majority of values centered between 5 and 6.

The continuous time transfer function of the second order system described by the differential equation above is

$$H(s) = 1 / (s^2 + 2\delta ws + w^2)$$

It corresponds to a z-domain transfer function (using pole/zero mapping)

$$H(z) = K/[z-\exp(-\delta w-jw\sqrt{1-\delta})][z-\exp(-\delta w+jw\sqrt{1-\delta})]$$

with K equal to

$$K = [1-\exp(-\delta w+jw\sqrt{1-\delta})][1+\exp(-\delta w-jw\sqrt{1-\delta})]$$

Discrete-time system representations were sought because of their usefulness for digital control. In Table 1, typically obtained transfer functions, H(z), are shown for one quadriceps electrode (for each patient).

Computer simulation was used to verify the experimentially determined models of the nonlinearity and dynamics). The models of Table 1 were simulated using LEANS (5) on an IBM XT computer. Experimentally obtained output data and data are compared in Figures 4 and 5. Similarly close correspondences were observed in all comparison trials.

In summary, it is possible by using fairly standard methods of classical control theory, to obtain a simple model of the electrode-muscle-joint system as a cascade of memoryless nonlinear gain and a second-order dynamic system, as in Figure 1, with a good fit between experimental observations and model simulations.

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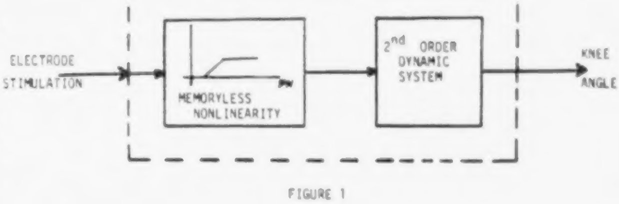

FIGURE 1

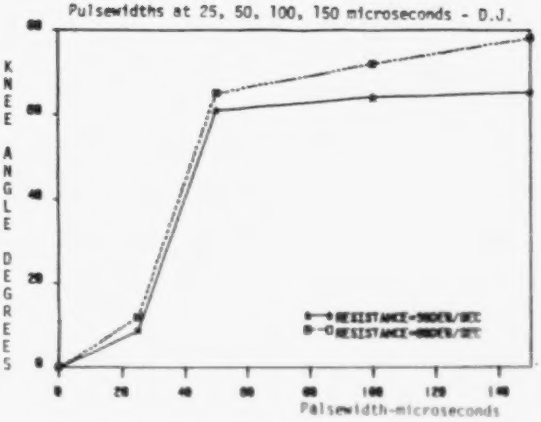

Figure 2

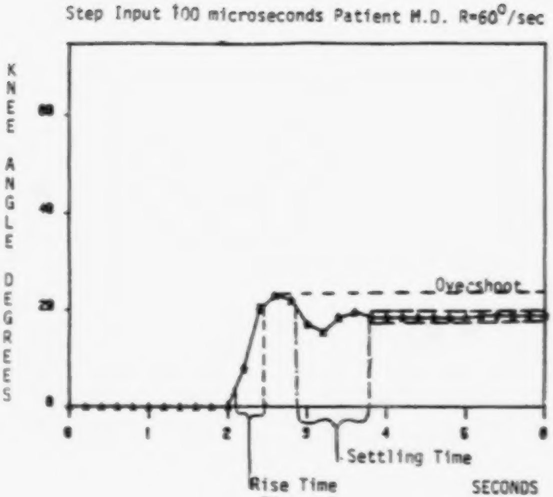

Figure 3

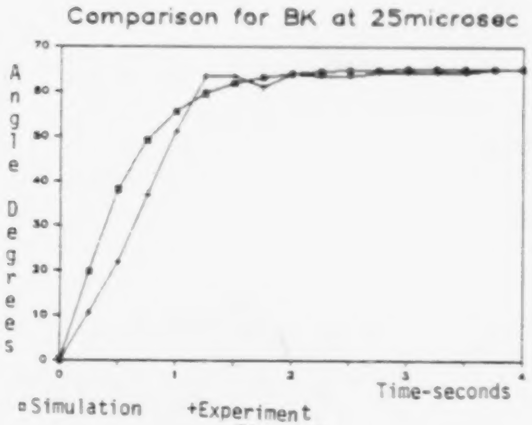

Figure 4

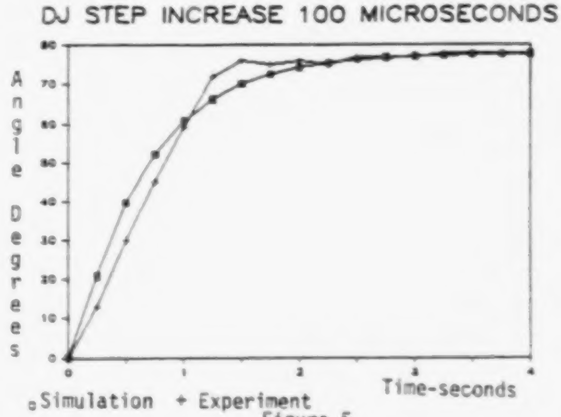

Figure 5

PATIENT	PULSE WIDTH - MICROSECONDS			
	25	50	100	150
MD	$H(z)=\frac{(1-.87e^{-j(.2)})(1-.87e^{j(.2)})}{(z-.87e^{-j(.2)})(z-.87e^{j(.2)})}$	$H(z)=\frac{(1-.92e^{-j(.2)})(1-.92e^{j(.2)})}{(z-.92e^{-j(.2)})(z-.92e^{j(.2)})}$	$H(z)=\frac{(1-.85e^{-j(.2)})(1-.85e^{j(.2)})}{(z-.85e^{-j(.2)})(1-.85e^{j(.2)})}$	$H(z)=\frac{1}{28.6z^2-40z+13}$
BK	$H(z)=\frac{1}{14.9z^2+22.4z+8.1}$	$H(z)=\frac{(1-.89e^{-j(.9)})(1-.89e^{j(.9)})}{(z-.89e^{-j(.9)})(z-.89e^{j(.9)})}$	$H(z)=\frac{1}{29z^2-42.9z+14.8}$	$H(z)=\frac{(1-.86e^{-j(.01)})(1-.86e^{j(.01)})}{(z-.86e^{-j(.01)})(z-.86e^{j(.01)})}$
DJ	NO RESPONSE	$H(z)=\frac{1}{27z^2-43.2z+17.1}$	$H(z)=\frac{1}{30.3z^2-42.7z+13.3}$	$H(z)=\frac{(1-.86e^{-j(.01)})(1-.86e^{j(.01)})}{(z-.86e^{-j(.01)})(z-.86e^{j(.01)})}$
SW	NO RESPONSE	$H(z)=\frac{1}{28.3z^2-45.6z+18.2}$	$H(z)=\frac{1}{45.8z^2-75.5z+30.7}$	$H(z)=\frac{1}{37.7z^2-57.8z+21}$

TABLE 1: TRANSFER FUNCTIONS - Z DOMAIN

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EXPERIENCE WITH A HELICAL PERCUTANEOUS ELECTRODE IN THE HUMAN LOWER EXTREMITY

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ABSTRACT

Helical percutaneous electrodes have been implanted in major muscles of the hip and leg of complete paraplegics. Preprogrammed stimulation patterns, delivered by a 32-channel microprocessor controlled stimulator, have allowed standing and functional walking. Analysis of the histories of 702 electrodes implanted in six subjects over a period of 29 months showed a probability of failure which decreased exponentially during the first four months, and then continued constant at 2-3%/month thereafter.

INTRODUCTION

A neuro-orthotic system to provide paraplegics with limited walking using functional electrical stimulation (FES) has been achieved in selected subjects with upper motor neuron deficit at levels T-4 to T-12 using surface electrodes [4], implanted percutaneous intramuscular electrodes [5], nerve cuff electrodes [1] and epineural electrodes [6]. These subjects have been able to walk with reciprocal, swing-through and swing-to gait using either a walker or crutches for support and weight transfer. The problem with surface stimulation has been the inability to provide the subject with selective stimulation of muscles, particularly the hip flexors.

Our approach has been to implant the subjects with temporary percutaneous intramuscular electrodes which allow selective stimulation of as many muscles as necessary to provide them with cosmetically acceptable and energy efficient gait. Once such a neuro-orthotic system is developed it can be replaced by a totally implanted permanent system using a total implant with either epimysial or nerve-cuff electrodes or a combination thereof. This paper describes our experience with implantation and retention of percutaneous intramuscular electrodes [2] in six paraplegic subjects over a period of 29 months.

METHODS

Only paraplegics with upper motor neuron deficit with level of injury between T4 and T12, and with no peripheral nerve damage, were considered for this study. They were given a physical examination and tested for availability of muscles amenable to stimulation. If qualified and accepted into the study, by their own consent, they were implanted with percutanous intramuscular electrodes [2].

TABLE 1

MUSCLES IMPLANTED WITH PERCUTANEOUS INTRAMUSCULAR ELECTRODES AND MOVEMENT FUNCTIONS

PRIMARY FUNCTION USED	MUSCLE	# IMPLANTED	% RETAINED
Knee Extension	Quadriceps	61	33
	Vastus Medialis	99	36
	Vastus Lateralis	72	43
	Vastus Intermedius	24	42
Hip Flexion	Sartorius	80	41
	Tensor Fascie Latae	50	52
	Gracilis	46	61
Hip Extension	Semimembranosus	60	36
	Gluteus Maximus	30	50
Hip Abduction	Gluteus Medius	24	33
Dorsiflexion	Tibialis Anterior and Peroneus Longus	66	53
Plantar Flexion	Soleus	63	36

Cadaver dissections were done to check for distribution of peripheral nerves to desired muscles and their access from the two points of insertion chosen bilaterally. One was on the medial aspect of the thigh, approximately half way between the hip and the knee to access hip extensors, flexors and abductors and knee extensors. The second point was on the medial calf just below the knee for access to ankle dorsi and plantar flexors.

The electrodes were made from 76-micrometer 10-strand stainless steel Teflon-coated wire. The wire was deinsulated at the end and wound over a .15-mm mandril. Approximately 10 mm of the coil and a hook were deinsulated at the end of the electrode and it was placed in a 19-gauge hypodermic needle. Implantation was carried out by a sterile procedure with all probe needles, wire electrodes and all the wire connections to the stimulator gas sterilized with ethylene oxide. First a point of maximal response to electrical stimulation was found with a 26 gauge probe near the motor point. The response was then matched with the needle carrying the electrode. The needle was withdrawn and the electrode left in place.

The electrodes were activated by a 32-channel portable microprocessor-controlled stimulator [7] which was programmed individually for each subject [5]. Muscles with implanted electrodes were stimulated cathodically with balanced charge, constant current pulses of 20 mA in amplitude, at a frequency of 25 Hz and a pulse width modulated from 0 to 150 microseconds to control the force of muscle contraction. The anodic reference electrode was carbon disposable dry gel (3M, St. Paul MN). The electrodes are covered with a Tegaderm (3M, St. Paul MN) film which lets water vapor out but is impervious to bacteria. The sites are washed daily with 70% isopropyl alcohol.

Between daily muscle exercise and thrice-weekly gait training sessions, the electrodes were stimulated from ten to fifteen hours per week. Electrodes were removed either because they no longer produced sufficiently large muscle responses; or because their measured impedance doubled; or because they produced an undesired muscle response. Selected removed electrodes were examined with a scanning electron microscope.

The histories of 702 electrodes were tabulated and analyzed using the life table and survival functions programs of the BMDP Statistical Software package (Department of Biomathematics, University of California).

RESULTS

Removed electrodes were observed to have an encapsulation plug around them at the point of entry into the skin, much like a hair follicle. The encapsulation appeared to extend deeply along the electrode, especially if it had been implanted for longer than six weeks. The electrodes were difficult to remove after six weeks or more; and they often broke on removal.
Electrodes which failed in the first three to six weeks after implantation were those which moved away from the nerve; these showed a normal impedance of approximately 1 KOhm but failed to produce the desired muscle response. The remaining electrodes generated repeatable and useful muscle forces. When mechanical failure occurred they showed a high impedance, and were no longer stimulated (Table 1).

Preliminary electron microscopy data of removed electrodes indicate the presence of some inclusions and manufacturing irregularities in some of the failed electrodes [3]. Corrosion had occured in some. Only five electrodes were removed for reaction suspicious of infection. On seven occasions subjects with positive cultures of staph aureus were treated with Keflex for 10 days, all with good results. Two subjects developed a generalized cellulitis of one leg which responded to antibiotic treatment. No specific electrode involvement could be found on clinical examination. One was found to have an infected pressure sore on his foot, unrelated to his participation in the project, which was identified as the cause of the cellulitis. No specific localization was found on the other subject. Both recovered completely with antibiotic treatment. One subject developed a superficial fungus infection in the electrode site on one leg which responded well to antifungal treatment.

The probability density function (Figure 1) showed an exponentially decreasing failure probability for the first six months followed by an approximately constant failure probability of 2-3%/month thereafter. The cumulative proportion of electrodes which survived is shown in Figure 2. No correlation was found between the survival function and either the subject or the muscle.

DISCUSSION

Because of the complex anatomical relationships and reliance on clinical judgement, the implantation technique was unpredictable, sometimes requiring many trials to implant a successful electrode. During the last month of the study the authors tested a new technique which they expect will permit more efficient implantations.

Fig.1: HELICAL PERCUTANEOUS ELECTRODE LIFE — CLEVELAND VAMC
PROBABILITY DENSITY FUNCTION

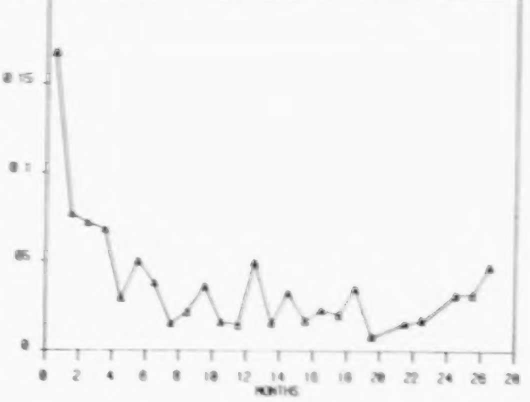
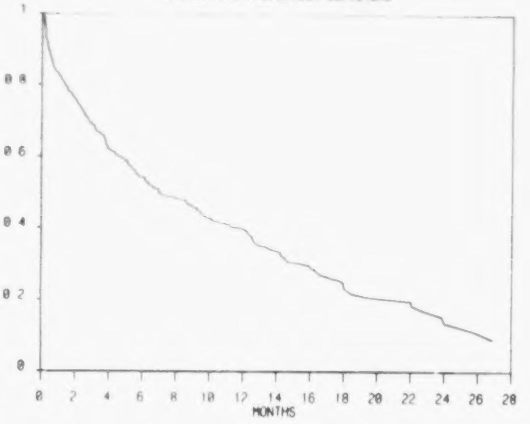


Fig.2: HELICAL PERCUTANEOUS ELECTRODE LIFE — CLEVELAND VAMC
CUMULATIVE PROPORTION SURVIVING



A disadvantage of using these electrodes is possible corrosion and resultant tissue damage should an electrode be stimulated after it has broken, and then deliver high density current. This problem has been minimized by weekly impedance measurement and discontinuing stimulation of electrodes which show high impedance. The incidence of infection has been kept low by adherence to sterile procedures during implantation and proper care of the electrode site.

The technique allowed selective stimulation of individual muscles to achieve specific functional movement in contrast to the use of surface electrodes with which repeatable, strong contractions of deep muscles are difficult to achieve. Further, the helical percutaneous intramuscular electrode provided an invasive, but reasonably safe capability for developing a neuro-orthotic system. The electrodes are very similar to widely used stainless steel suture material which has been safely used in the human for many years.

Selected X-ray evaluations of broken remnants of this type of electrode have revealed no evidence of migration after periods as long as twenty years.

Possible future improvements in the electrode design include better quality control of the wire as it comes from the manufacturer, increasing the number of strands, winding two electrodes in tandem or using a stronger material.

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16.1

HOW TO BEGIN A MICROCOMPUTER PROGRAM

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ABSTRACT

This paper describes the development of a microcomputer program with physically/mentally handicapped students at a residential hospital/school. The microcomputer program has added significantly to the education and quality of the student's lives.

INTRODUCTION

New technologies have been developed in conjunction with the microcomputer that allow severely handicapped individuals the access to interact with life, they no longer must be passive observers. Crippled Children's Hospital and School has developed a program to utilize these new technologies with their severely handicapped students.

BACKGROUND

Crippled Children's Hospital and School (CCHS) serves a school population of approximately 120 handicapped children who range chronologically from ages 4 to 21 years and developmentally from 1 month to age appropriate developmental levels. The students exhibit a variety of diagnoses including: Cerebral Palsy, Muscular Dystrophy, Spina Bifida, Seizure Disorders, Post Trauma, and Ataxia Telangiectasis. The school structure incorporates pre-school through high school curriculum levels. Students range from mildly to moderately to severely physically handicapped, verbal to non-verbal, and intellectually from above average to severely mentally impaired.

The focus of this paper is to relate how Crippled Children's Hospital and School implemented a microcomputer program; staff involvement; hardware and software selection; adaptations and student impact.

BEGINNING A PROGRAM

In May of 1982, funding was received through a grant from the South Dakota Department of Education to purchase 5 microcomputers and a selection of devices for accessing them transparently. A committee of special educators and speech therapists were in charge of selecting the appropriate hardware and software.

One speech therapist and one special educator from this committee attended a workshop on the use of the microcomputers at Trace Center, University of Wisconsin-Madison. Here, they obtained a basic knowledge of different types of available microcomputers, use of microcomputers, and the available adaptations for the handicapped. These two staff members were then put in charge of further purchasing and maintenance of the computer program.

BASIC SET-UP

For our CCHS program, the Apple IIE computer was selected because of it's already widespread educational use and availability of communication software packages. At the time it was also the only microcomputer with which the Express III and Zygo 100 would interface. In addition to the Apple IIE, the following peripherals were purchased: printer, disk drive, color monitor, and numeric keypad.

Computer stands with wheels were constructed for the above mentioned basic set-ups. Due to the differing heights of the many electric and manual wheelchairs, the stands were constructed to be raised and lowered using lead screw assemblies (a threaded rod with a follower such as a nut on it) from manual hospital beds. This has proven to be beneficial.

Four of the five microcomputers were then placed in the classrooms. One was placed in the Non-Verbal Augmentative Communication Evaluation Center for use during speech therapy and occupational therapy sessions and for evaluation purposes. The committee felt these placements would utilize the microcomputers to their optimum potential.

HARDWARE

The following is a list of the hardware we purchased with the grant money. Listed are a few advantages and disadvantages we have found at CCHS. These are our observations from the situations we have experienced.

1. Numeric key pads
   Advantages - much software requires only numeric input; keypad can be placed on student's laptray and student need not be repositioned; ziplock storage bag can easily house keypad and protect it from drool and spills.
2. Keyguards
   Advantages - our maintenance department constructed 1/2" thick clear plexiglass keyguards which are cheaper and more reliable.
   Disadvantages - commercially available keyguards are too expensive; often interfere with computer function.
3. Express III interface
   Advantages - allows an Express III user (with any switch) to interface with an Apple computer and utilize the computer through their machine.
   Disadvantages - if child is using different computers in different settings you would either have to have additional interfaces or transport that one around.

4. Zygo 100 interface
   Advantages - allows Zygo 100 user with any switch to interface with an Apple computer and utilize the computer through their machine.
   Disadvantages - same as Express III
5. Shadow-VET
   Advantages - allows quadriplegics and post trauma victims to run the computer using voice input.
   Disadvantages - not easily used with our cerebral palsied population due to modulations of their voice, inconsistent articulation and lack of breath support.
6. Tetra-scan
   Advantages - allows the use of many different switches.
   Disadvantages - difficult to install cables
7. Adaptive Firmware
   Advantages - easy installation; allows many different accessing modes; allows use of different switches
8. Echo II Speech Synthesizer
   Advantages - produces understandable synthetic speech; adds audio for a more complete input for a student; easy installation.
   Disadvantages - not loud enough for some students.

SOFTWARE

The next step was to find appropriate commercial software for a diverse population. The committee found that standard educational software can be modified for use with special needs children if a small amount of teacher time is utilized.

With the passage of time, more software targeted for preschool and mentally retarded students has become available. We have found software which allows for the setting criterion, changing scanning rates, changing response times and allowing for different input modes to be the best choice for the physically/mentally handicapped student.

The following is a list of basic software purchased with the grant money. It is not a comprehensive list of all software at CCHS, but gives a feel for what we have here. Once again, there are advantages and disadvantages we have observed here. (All are available through dealer catalogues or at a computer store.)

1. First Words, First Categories, MICRO-LADS (language and basic vocabulary)
   Advantages - color graphics and animation; synthetic speech with the ECHO II speech synthesizer; offers audio/visual rewards for correct answers; offers different levels; offers testing of knowledge gained; data print-out sheet available.
   Disadvantages - children with auditory perceptual deficits have difficulty understanding the speech.
2. Talking Blissapple (Blissymbolics)
   Advantages - allows Bliss user to write in Bliss with the use of a printer; allows for black on white or white on black symbols; has voice output with Votrax voice synthesizer.
3. Early Games (preschool concepts)
   Advantages - colored graphics; accompanying sounds; selection of games without reading ability.
   Disadvantages - no data collection; does not allow for lower case letters.
4. GULP (math)
   Advantages - records score on screen so children can compete with one another; must first get math facts correct before game can begin; non-threatening way to learn math.
5. MATH WARS (math)
   Advantages - allows learning of math in game setting; teacher can select level and exact skill student works on.
6. Delta Drawing (drawing and early programming)
   Advantages - allows handicapped children to draw with the computer; allows children to learn about keyboard and simple programming in a non-threatening way; easy to make changes in pictures.
7. Spelling Machine (spelling)
   Advantages - data keeping system; built in reinforcement system; color graphics and sound.
   Disadvantages - takes time to add words not in software's dictionary.
8. Grab a Byte (nutrition and home-ec)
   Advantages - reinforces basic nutrition concepts taught; allows for students to plan a balanced meal.
   Disadvantages - does not include enough foods.

IMPACT ON STUDENTS

The instructors feel the students are more motivated to learn with the use of the computer as an aid. Students have self-directed activities which do not require instructor's time, students have a non-discriminating "teacher" who will do drill and practice without offering negatives or judgements.

Independence, one of the best "rewards" the computer has given our students. For the first time in many of their lives, students can independently write a letter, do a math paper, play a game, or offer vocational choices.

CONCLUSION

Designing a program for the use of this technology can be frustrating and frightening due to rapidly changing technology, lack of knowledge and lack of funds. In developing a program, remember the following three things:

1. Many pieces of hardware and software can easily be purchased through catalogues and at computer stores. They provide a stable base upon which a computer program can be built.
2. Most hardware and software comes with understandable, step-by-step manuals.
3. There is an increasing number of resource people in the community to assist and perhaps develop your program with you.

Maintain an open mind and explore all available possibilities. Often the most unexpected thing may be the key for the person you least expected.

Plunge in, learn the technology as you go and everyone will profit from it.

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THE ELEMENTARY MOD KEYBOARD

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ABSTRACT

A software module in the form of a plug-in cartridge has been developed for the National Research Council of Canada MOD Keyboard System, which is a dual-computer system. This particular module is intended for severely physically disabled students who require large letters, clear presentation of items, and access through 1 to 5 inputs. The module allows an instructor to create and store, on the cartridge, a series of front-end displays tailored to various applications. To facilitate use by multiple students individualized user parameters, such as scan interval, may also be defined and stored. A clinical evaluation of this system has been carried out and a number of display design guidelines have been delineated.

INTRODUCTION

Originally developed by the National Research Council of Canada, the MOD Keyboard (1) allows a disabled person who is unable to use a conventional keyboard to enter information into a computer and use unmodified commercially-available software. A dual-computer approach is employed with a VIC-20 as the front-end computer. With the Elementary MOD Keyboard (EMK) all standard keyboard functions may be entered using 1 to 5 inputs.

The EMK works on the principle of replacing the normal keyboard with a second, front-end microcomputer whose screen displays a customized keyboard arrangement. Items on this screen are scanned with a cursor controlled by input switches. When the cursor reaches a desired item, a switch is pressed and the item selected is transmitted via a cable to the main or back-end computer which interprets it as if typed at its keyboard. Thus, a disabled user may type without ever touching the back-end computer's keyboard.

The computers with which the EMK may be used as an alternative keyboard include the Apple][+, //e, IBM PC, Commodore 64, or any other computer that has the facility to accept keyboard input from another computer.

MAIN FEATURES OF THE EMK

The foremost feature of the EMK is its flexibility and its easy adaptation to a wide range of users and software applications. **Items** displayed on the front-end screen may consist of characters, words (maximum 20 characters), or phrases (maximum 20 characters). Each of these items may also be defined to represent different or expanded **messages** (maximum 60 characters). Items may be positioned anywhere on the screen. A maximum of 8 **pages** or screen arrangements of items may make up a "keyboard-on-the-screen" or **display set.**

Display sets are created or edited by an instructor to suit any particular software application. For example, when using a math program, only the numbers and operation symbols required for the program may need to be displayed. This feature allows the student to focus attention on selecting the important items, rather than becoming frustrated or distracted by unnecessary or confusing items. Two **character sizes** are available, the **standard** VIC-20 characters, and an **expanded** size (characters which are twice the regular height, but the same width).

User parameter sets are also created or edited by the instructor to define how a student controls the MOD Keyboard, for example, **selection strategy** and **scan interval.** Student input may be through 1, 2, 3 or 5 switches, using element, row/column or directional selection strategies.

System parameters may be defined to adapt the EMK to various back-end computers. Both serial and parallel output is supported as well as a programmable time delay between transmitted characters. This time delay is particularly useful when the back-end computer cannot buffer characters or respond fast enough to a string of characters.

The **EMK cartridge** has a built-in memory (8K of battery backed-up CMOS RAM) that can store a maximum of 32 different user parameter and display sets.

Three selection strategies are available:

Element scan (1 or 2 inputs): the cursor moves automatically from one item to the next. Items are scanned from left to right, top to bottom. A quick activation of one switch selects the desired item when the cursor highlights it. Maintained activation of a second switch advances the cursor at a faster rate to quickly move through the display items.

Row/column scan (1 or 2 inputs): the cursor automatically highlights rows of items starting from the top and moving to the bottom. The row with the desired item is selected through a quick activation of one switch and then the cursor moves across items in that row. Another activation of the same switch selects the desired item when the cursor highlights it. A second switch can be used to allow the student to switch the cursor from scanning items to scanning rows if that particular row was mistakenly selected.

Directional scan (3 or 5 inputs): the cursor moves up, down, left or right on the screen to the desired item through maintained activation of the joystick or a set of two (right and down) or four (up, down, right and left) switches until the desired item is highlighted. A quick activation of another switch selects the item.

OPERATION OF THE EMK

When the EMK is turned on, a title/credit page will be displayed for several seconds before advancing to the second page which displays a list of user parameter sets. A user parameter set may be selected via the VIC-20 keyboard or via a single switch by the student. This action will advance the user to the display set page where selections can be made in the same fashion. The selected display set will now be operational according to the specifications in the chosen user parameter set. The student may independently return to each of these previous pages to select alternate sets. User parameter and display sets may be edited and deleted from the respective pages by an able-bodied instructor.

CLINICAL APPLICATIONS OF THE EMK

During its development the EMK was evaluated with nine physically disabled students to determine its effectiveness in practical applications and its ease of operation for both the student and the instructor. The information derived from the evaluation was integrated into the final version of the software and the User's Manual.

All the students involved in the evaluation were severely physically disabled and required special accessing methods. The students ranged in age from 12 to 28 years. Only two students were proficient readers and writers, the others were beginning to learn to read and write. All but one student had little or no functional speech. Interest in and ability to use a computer varied among the students.

Student Use

All selection strategies, except the 5-switch directional scan, were used in the evaluation. An appropriate strategy was chosen for each student according to their individual abilities. Initially when a new display set was introduced, the scan interval was set longer than the optimum. Once the student became familiar with the display set and software program, the scan interval was shortened to a rate which matched his/her physical ability. The ease of quickly altering user parameters made it convenient to try out different strategies and to tailor the strategies to the individual student.

None of the students had difficulty in learning to use the dual-computer system. Difficulties that arose were related to the students' physical accessing and attention. Students displayed no difficulty relating to two monitors, often placed side-by-side, angled inwards about 30 degrees. One student who had left-side neglect found it easier to use the EMK when the monitors were set one on top of the other.

While the student can be independent in selecting user parameter and display sets, these sets must be created and entered by an able-bodied person through the keyboard. This was not deemed a significant problem for the population that the EMK is addressing. In the evaluation it was found to be beneficial to involve the students in the determination of screen items and to observe the instructor entering the information to gain a better understanding of the EMK.

Instructor Use

The software and User's Manual are complementary and have been designed to be straightforward and easy to use. A tutorial in the manual takes a new user step-by-step through the main features of creating user parameter and display sets. A detailed reference section serves to supplement the user's knowledge in order to exploit the full capabilities of the EMK. The evaluation demonstrated that the workings of the EMK can be learned in a 1 to 2 hour period. The development of display sets which will facilitate efficient access of items by the student, however, takes more experience in using the system. Appropriate labels and arrangements of items must be considered in attempts to minimize the number of switch activations and the time required to scan items.

Display Sets

Arising from our experience with the EMK and other transparent keyboard systems such as the adaptive-firmware card (2), principles regarding the set-up of screen displays have been developed and are continually being evaluated.

The EMK allows for the creation of displays which are specific to particular software packages. Thus, the student does not have to contend with unnecessary items which may be visually confusing or distracting.

A display set is developed with the following factors in mind: requirements of the back-end software, selection strategy to be used, and communication level of the student(s). Different display sets may be created to satisfy the needs of students who are functioning at different levels.

The expanded letter option was one of the most highly rated features of the EMK since many students had difficulty seeing the regular-sized letters as typically displayed on the Apple computer.

The paging function is a useful option for those students who can benefit from the flexibility of larger and more powerful display sets. The first pages should contain the frequently used items and the latter pages reserved for less frequently used items. An accompanying reference display, listing all pages of items, placed beside the EMK monitor was found to be helpful for the student when using display sets of more than two pages .

There are a number of ways of displaying the alphabet that may be more appropriate for each student. The traditional "A to Z" arrangement may be a logical one for a student who is just learning the alphabet. The QWERTY arrangement may be appropriate for a student who is familiar with the layout of the typewriter keyboard. Efficiency of access can be facilitated through the use of a frequency-of-use arrangement of letters.

For the element and row/column selection strategies, frequently used characters (space, backspace, return, next page) are best placed in the upper left-hand corner of the screen where they are scanned first and thus accessed quickly.

Text editing display sets with "columns" of words and phrases (i.e. subject, verb, object) were found to be more effective than those with separate pages of words ordered according to parts of speech.

Another effective text display for many users consisted of a vertically split screen with the letters of the alphabet in five columns on the left of the screen, and a collection of words or abbreviations on the right (Figure 1). Each abbreviation can be defined to transmit a message of up to 60 characters in length. Several pages can be created in this fashion and the user can access both the alphabet and their abbreviations without having to shift between the two pages.

The ability to redefine items on the screen is a very powerful function. For commonly used phrases, coded input, and sequences of commands, a redefined message can save the user a tremendous number of interface activations, and hence time and effort. Also, the item on the screen can have a more meaningful representation than the message itself. When redefining words, or phrases it is often helpful to include a space following the last word since this is the next most likely entry.

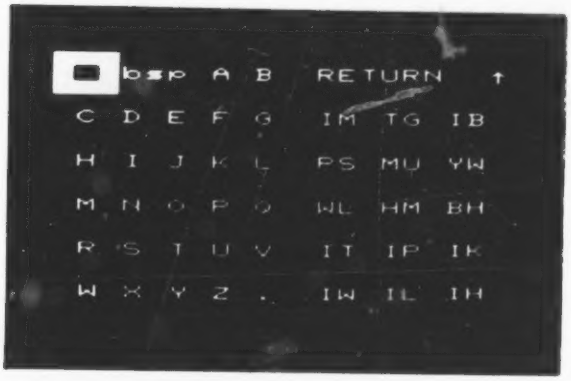

Figure 1: EMK display set illustrating complete alphabet and abbreviations.

The flexibility in the creation of display sets allows for positioning of directional commands (up, down, left, right) in a spatially logical pattern. For the younger student this approach reinforces the meaning of the selected response.

Simple clear acetate overlays greatly enhance the EMK's utility, particularly for non-reading students. For example, arrows rather than words can be placed in the appropriate positions on the acetate overlay. The screen items behind these arrows would be redefined as the directions required for the back-end software. Pictures or Blissymbols can also be put on overlays and the blank screen items behind them redefined appropriately. The cursor will still scan the blank spaces and highlight the items on the overlay. Caution should be taken to ensure that the back-end software is complementary to the student's comprehension and communication level.

CONCLUSIONS

The EMK has proven to be a very powerful front-end computer system that allows severely physically disabled students to access the same software used by able-bodied students. Its prime advantage is that it is readily and easily tailored to both the student's abilities and the requirements of the software application. The free-format display layout allows an instructor to create an application-specific display from which meaningful items can be selected without distraction from confusing and redundant items. The abbreviation/expansion message feature also minimizes the number of inputs required.

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16.3 USE OF COMPUTER PROGRAMS TO TRAIN SWITCH ACTIVATION SKILLS WITH YOUNG CHILDREN WITH HANDICAPS

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ABSTRACT

Single case designs were used to examine the effects of specific intervention strategies used with nine young children with multiple handicaps. Three levels of training were provided: (a) single switch activation; (b) discriminated activation; and (c) pre-communication training.

A Commodore Vic-20 computer was used to deliver the programming cues and consequences and to collect data on child performance. Two primary software programs implemented time-based and trial-based training programs. Training was provided for all children by their teachers and within the classroom. Results indicate that training can be easily and cost effectively delivered through the child's educational program and with effective results.

INTRODUCTION

Many young children with severe and multiple handicaps must learn to activate switch interface devices in order to control aspects of their environment. Environmental control is not only essential to performance of leisure (play), mobility, and communication skills but may also provide a means for furthering the cognitive and intellectual development of the very young child (1). The use of environmental control devices can be facilitated through systematic training that focuses on teaching the child the cognitive skills necessary to use existing movement patterns for switch activation (2,7). Previous efforts by these investigators have demonstrated that even adolescents with severe physical handicaps and significant cognitive impairment can learn to activate switches for functional use if trained with precise intervention methods (3,4).

BACKGROUND

Providing training in switch activation for children with significant cognitive impairment has been suggested by a number of investigators but systematically tested by only a few. Most training systems have depended on use of switch interface systems hooked to battery operated consequences. Child performance data, if collected, has often been obtained using either a live observer or an electro-mechanical counter. Establishing time-based training strategies have required hand timing by an independent observer (e.g., 3, 4, 6). Computer systems that have been used have required costly investments for both hardware and software and have still required observers to collect the required types of data (1). Providing switch training using any of the existent systems has been costly both in terms of necessary equipment as well as personnel time. Most programs that have been reported have required specialized staff to be able to implement switch interface training programs.

Very young children function intellectually within the sensorimotor period of development. Children learn to use movement abilities in successively more complex organizational patterns through interacting with objects in their environment. The initial movements demonstrated will be increased in terms of rate of performance when consequated by a novel or interesting event. Consistent exploration of objects in the environment provides the opportunities for children to learn to perform movement skills differentially or in discriminated ways. As such, a movement pattern that was initially performed under the control of the consequences to that movement becomes performed under the control of the stimulus event(s) that precede the movement. Manipulation of consequence events can be used to increase the performance of a movement that is within the repertoire of the child but performed at an insufficient rate to be functional. Antecedent cues, prompts, and other strategies can then be implemented so as to shift the movement pattern to the control of stimulus events (4,6). Performance of one (or more) movement patterns under the control of stimulus conditions is essential to all types of activation of switches in various functional activities. However, this level represents only a minimum level of performance. Ideally, an individual must be able to chain together two or more discriminated movement patterns in order to be able to interact with the environment in functional ways.

MATERIALS AND METHODS

A Commodore Vic-20 computer was modified with the Vic-Rel input /output device to deliver one of two programming methods for children participating in this investigation (5). The computer set up was located within a classroom for toddler and preschool aged children with a variety of handicapping conditions. A range of switch interface devices were available. These individually fabricated or commercially purchased switches were matched to the motor dysfunction of the child so as to match the most frequently performed movements with the switch interface that was most easily operated by the child. Various types of consequences were also available and included: special toys; taperecorded music or events such as the parent reading a story; and light configurations. Consequence hierarchies were obtained for each child using the available consequences. These hierarchies were determined on the basis of highest to lowest rate of activation, using the most easily activated switch interface.

Training Method 1

Software was developed that allowed programming for single switch activation and for discriminated activation of two similar switches. The software was designed to: (a) provide the teacher selected consequence when the appropriately cued switch was activated; (b) monitor the rate of activation of either the single switch or the "live" switch in discriminated training and provide an additional consequence if the rate deviated from a rate pre-established by the teacher; and (d) report data concerning rate of activation of both "live" and "dead" switches and number of consequences provided. The programming was set up such that each child received two 10 minute training sessions per day. These sessions were further divided into two five minute segments. The position of the "live" and "dead" switches were held constant throughout one 5 minute segment and were counter balanced across the 20 minutes of training such that the right position switch was "live" for two 5-minute segments and visa versa.

Training Method 2

The second training strategy was designed to allow for counterbalancing of the "live" switch on a trial-by-trial basis. This option was created so as to allow for randomization of switch placement across a predetermined number of training trials. Software was developed to: (a) create the randomization of "live" and "dead" switches; (b) allow for a random delivery of up to 6 different consequences (predetermined by the teacher); (c) provide variable length of consequences for "live" activation; and (d) summarize data at the end of the training session. Training session length was also variable but was established prior to the start of the session (and was dependent on the number of training trials provided.)

Methods Used

Three children were initially programmed for with single switch activation using Training Method 1. Children who were trained to activate switches discriminatively used both Training Methods 1 and 2. Children in precommunication training, in essence, were performing discriminated activation but were required to activate in relation to named pictures that were placed on the switch interface devices. In part, Training Method 2 provided the best option for children with significantly different rates of activation in the right and left upper extremities.

Single case designs were developed for each of the children. Single switch activation was tested using reversal designs. Discriminated performance was tested using: (a) multiple baseline designs across different types of switch configurations; (b) alternating treatment designs contrasting Training Methods 1 and 2; and (c) reversal designs. Data were maintained on a session by session basis and used to make programming decisions concerning adjustments in training.

Subjects

A total of nine children with severe and multiple impairments participated in this training. The mean age was 2.6 years with a range of from 13 months to 3.0 years. All children showed significant developmental and cognitive delays. None was verbal or ambulatory. All had significant deviations in movement abilities.

RESULTS

All participating subjects demonstrated acquisition of switch activation with at least one switch. The six children who were trained in discriminated switch activation were successful in bringing existing movement responses under the control of antecedent stimuli cues. This included the three children who received precommuncation training who were required to discriminate picture cues.

DISCUSSION AND CONCLUSIONS

The data available from these nine children indicate that young children with severe and multiple handicaps can acquire the movement and cognitive skills necessary to activate switch interface devices for functional purposes. These results replicate previous research (1,3, 4,6) that has demonstrated benefits from similar types of training for both less handicapped children and older (adolescent) students. The significance of this investigation may lie more with the ease of replicability that is possible and that has not been easy to manage with previous studies. This training system could be repeated by teachers and parents, with limited training, using the low cost hardware and software developed for this investigation.

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16.4

AN INEXPENSIVE BAR CODE READER SYSTEM FOR OBTAINING INFORMATION FROM "PRINT"

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ABSTRACT

A large number of persons with disabilities are unable to obtain information from the available symbol systems. Although a variety of technologies are available for solving the problem, they are extremely expensive or require considerable training and skill to use. A simple inexpensive alternative is the use of bar code reader technology. Bar code readers transcribes printed bar code into synthetic speech. Applications of this technology in the areas of communication, Braille training, and self-instruction are discussed. In addition, design limitations and recommendations are presented.

INTRODUCTION

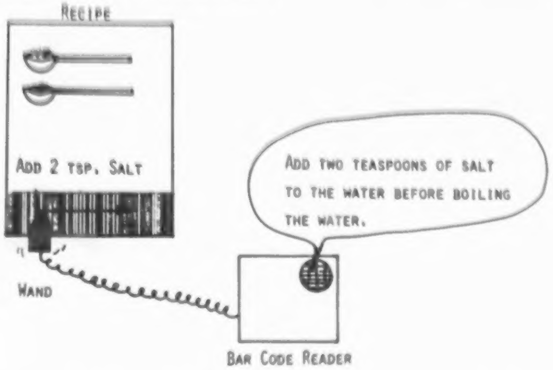
A large number of individuals with handicaps, such as mental retardation and visual impairment, are functionally illiterate in regards to gaining information from the available symbol systems. Without skills to use a symbol system, many persons will forever remain dependent upon others for their information needs. Also, since most self-instructional aids, such as cookbooks, first aid manuals and board game rules, are designed to be read many individuals with handicaps are unduly restricted in regard to personal growth and independence.

Currently there are a variety of devices and systems that enable persons with disabilities to obtain information. These devices and systems include Braille, pictorial instruction (8,10), Talking Books (5), the Optocon (7), and the Kurzweil Reading Machine (4). Each of these alternatives is successful in serving many of the informational needs of a large number of individuals. However, as a result of more severe learning impairments and/or the expense of the technology, the currently available alternatives do not serve the needs of many persons with disabilities.

Bar Code Reader Technology

Another strategy for effectively obtaining information from a variety of sources is the use of bar code reader technology. For many years this technology has been used to obtain reliable information for merchandise inventory and sales. Basically this system involves passing a light sensitive pen or laser over a printed code containing bars of varying thickness which correspond to coded numbers. This bar code can be printed anywhere on an item. Information regarding the product that is stored in the machine's memory can then be accessed and presented via a visual display, printed on paper, or spoken aloud with the aid of a voice synthesizer.

An information system for persons with disabilities would consist of a portable device which could "read" printed bar code from a variety of sources. The processing unit would then transcribe the code and auditorily present the information via a speech synthesizer.


RECIPE
ADD 2 TSP. SALT
ADD TWO TEASPOONS OF SALT TO THE WATER BEFORE BOILING THE WATER.
WAND
BAR CODE READER

A bar code reader system could be effectively used by individuals with a broad range of disabilities, from mentally retarded persons to severely visually impaired persons. This system would have a wide variety of uses as a means of obtaining new information and retrieving forgotten information.

Requirements of a Portable Bar Code Reading Device

In order to meet the information needs of disabled individuals, a decoding device must satisfy at least the following requirements:

1. The device must rapidly and reliably decode and transmit the desired information.

2. The device must be low cost, averaging less than approximately $60 per unit so that they may be purchased by disabled individuals, medical insurance companies, and public funding sources, as are other necessary prosthetic devices such as eye glasses and hearing aids.
3. The device must be widely commercially available or be easily constructed by any competent electronics technician using readily available components.
4. Since disabled individuals would use the devices in a wide variety of environmental settings, the devices must be portable. Portability involves five separate factors. The devices must be: a) light weight and easily hand-held; b) able to be carried in a pocket, worn on one's belt, or hung from a shoulder strap like a personal radio or cassette player; c) able to operate away from conventional power sources for at least one day (i.e., twelve hours); d) robust and durable enough to survive the wear and tear of daily use in vocational, educational, residential, and community environments; and e) operate correctly under a wide variety of climatic conditions.
5. As the device will be used in a variety of environments (e.g., work settings, public transportation, homes, restaurants, schools and recreational facilities), the device must be unobtrusive and readily accepted in those settings.
6. The device must be easily operated by individuals with a wide variety of disabilities.
7. The device must produce words and sentences clear enough to be understood by normally hearing individuals.
8. The device must also be flexible enough to accomodate future changes in information needs of disabled individuals.

Magic Wand Speaking Reader

Texas Instrument's Magic Wand Speaking Reader is the only commercially available device that appears to meet most of the aformentioned requirements. The Magic Wand "reads" bar code via a hand-held light sensitive wand by passing the wand over the bar code. The processing and speech synthesis unit then produces the corresponding words or sounds. The speech can be sent to either a speaker (e.g., this will be useful when following a recipe) or to an earphone (e.g., this will be useful in a store or work setting). At approximately $47.00, this device is low cost and it is available at thousands of commercial outlets across the U.S.A. In regards to portability, it is light weight, operates on batteries or AC current, it is robust and durable, and it can operate under a variety of climatic conditions. Although the Magic Wand originally comes in a 20cm bright blue disk it can be repackaged into a box as small as 4x10x8cm. The Magic Wand is an extremely flexible tool that is capable of producing a virtually unlimited number of letters, numbers, words and sentences, multiple sounds, sound effects and music.

CURRENT DEVELOPMENT ACTIVITIES

Research and clinical applications of bar code readers as aids for persons with disabilities fall into three basic categories. The first application area is in interpersonal communication. This involves using the Magic Wand as an inexpensive voice output communication device (1), and as a device for speech remediation (6) Work is also currently underway to develop a comprehensive communication system called "Tigerbook" for use by persons with disabilities(2).

The Magic Wand is also being used for teaching Braille (9). Books produced for the Magic Wand were adapted for Braille instruction by simply placing Braille over the text and adding textual symbols to make tracking easy. This system enables students to study Braille by themselves and it frees the teacher to work on other activities.

The third area involves using the Magic Wand for obtaining new information and retrieving forgotten information.

Some of the uses of the Magic Wand as an informational tool include:

1. Enabling severely visually impaired persons to "read" a restaurant menu equipped with bar code.
2. Providing instructions on how to place emergency calls including whom to call, how to call, and what information to give.
3. Enabling nonreaders to "read" bus and recreational actively schedules.
4. Providing instructions (bar coded recipes) on preparing nutritious meals.
5. Providing instructions on how to perform a wide variety of tasks such as completing a checkbook, doing laundry, using a kitchen tool, playing a board game, treating emergency injuries, and performing a new work task.
6. Providing instructions on how to properly take medication.

Bar codes can be used to supplement virtually any instruction or information that can be provided in a printed format. A research project directed by the author is currently evaluating the Magic Wand as a reader for the above mentioned purposes.

Design Needs

Although the Magic Wand has proven to be valuable for a variety of purposes, several concerns have been identified in regards to its design and performance. One problem is the limited vocabulary that is currently available. The "Talking Tracks' (bar codes) which are read by the Magic Wand were produced by Texas Instruments from two to three years ago. A system for rapidly producing customized codes to meet individuals' needs must be established before widespread use of the Magic Wand can occur.

A second limitation is the precision required for properly tracking the bar code. The problem of precision causes difficulties for young users and for users with physical impairments or severe intellectual impairments. Modifications of the wand to facilitate grasping, tracking guides, and careful positioning of bar codes might ameliorate some of the tracking problems.

Although the Magic Wand's speech quality is generally acceptable, numerous words are still unintelligible. From the time when the Magic Wand was designed to today there have been considerable improvements in synthetic speech production. These advances should be incorporated into a redesigned bar code reader.

The final concern involves the Magic Wand's power consumption. Although the device can be supplied by power from an AC adapter, batteries are used for most applications. Depending upon the intensity of the use and the types of batteries used, the batteries last approximately one to four hours of constant use. Incorporating advances in electronic technology, such as CMOS speech synthesizers (3) into a redesigned bar code reader should greatly reduce the units power consumption.

CONCLUSION

The bar-code reader system should prove to be a time-efficient procedure for promoting self-control and independence. The Magic Wand provides auditory stimuli which persons with disabilities can use to guide their performance across diverse settings, tasks, and time. In addition, the bar-code reader system is potentially efficient because: a) it is easily used in new tasks and settings; b) it is easily adapted to meet differing information needs; and c) its' use will likely produce a reduction in training time. The use of bar code readers as efficient, self-operated information and communication devices deserves careful development and evaluation. Bar code readers have the potential for increasing the independence of a large number of individuals with disabilities.

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WORKERS' COMPENSATION
AN OPPORTUNITY FOR REHABILITATION TECHNOLOGISTS

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ABSTRACT

Workers' Compensation is a benefit provided to thousands of injured employees across the country each year. Historically, this benefit has included vocational rehabilitation services. These services have tripled in cost since 1978, are less effective than modified or alternative work, and have not significantly impacted the injured worker's income.

Most Workers' Compensation clients are only mildly to moderately impaired. Approximately two-thirds of the injuries involve either the back or the upper extremities. Many of the injured workers could be accommodated through rehabilitation technology, thus allowing return to original or similar work.

A technology-oriented fee-for-service enterprise has been functioning within California's Workers' Compensation system since 1976. It is based upon the objective determination of the needs through a functional job analysis including occupational biomechanic, work place design, and environmental factors. This problem-solving approach and its results are discussed in detail.

BACKGROUND AND NEED STATEMENT

"Make as few changes as possible in the disabled employee's life. Modified or alternative work - a substantially similar job in a substantially similar workplace, frequently with substantially similar co-workers - provides the best chance for successful rehabilitation. Other, more complex training cannot match the cost-effectiveness of short-term rehabilitation programs that minimize the upheaval in the worker's environment." This quote is from a 1983 publication, A Report to Industry: Vocational Rehabilitation, by the California Workers' Compensation Institute (CWCI) and clearly states a conclusion drawn from the results of ten years' experience.

California was the first state to mandate vocational rehabilitation as part of their Workers' Compensation benefits (1974). Consequently, California has the largest record with which to track service delivery trends. The following statistics are taken from the same CWCI report quoted initially.

Cost to Industry

1. Estimated 1982 cost of vocational rehabilitation to industry in California was $130,000,000 ($130 million).
2. Estimated % cost to employers for vocational rehabilitation benefit:
1976 - 2.7%; 1980 - 7%; 1983 - 10% (est.)

Cost of Service

Vocational rehabilitation average cost figures include maintenance allowance, evaluation, testing, training, and retraining: 1978 - $ 4,218; 1980 - $ 6,927; 1982 - $11,000 (estimated); 1983 - $13,000 (estimated).

Duration of Service

1. Modified or Alternative Work - 100 days
2. Formal Schooling - 226 days

Rate of Success

1. Modified or Alternative Work - 85%
2. Formal Schooling - 52%

Rate of Pay After Vocational Rehabilitation

Lower than at injury - 46%
Same as at injury - 9%
Higher than at injury - 45% (does not consider raises that would have occurred).

It is surprising to many people, particulary the injured worker, when they discover that Workers' Compensation vocational rehabilitation is not their way to becoming a doctor, lawyer, or computer specialist. Also surprising is the fact that the employees who are eligible for and request the vocational rehabilitation services are generally not the more severely disabled. Within the workers' compensation system, it is the less severely disabled who generally request the service as seen in the following chart.

% of Permanent Disability	% Eligible	% Accepting
≤ 10	13	86
11 - 24	25	63
25 - 49	36	72
50 - 69	52	61
≥ 70	71	50

This situation occurs for a variety of reasons. One is that many Workers' Compensation carriers and counselors consider the severely injured (disabled) as "nonfeasible."

The majority of workers requesting vocational rehabilitation have injuries to either their backs (one-third) or to their upper extremities (one-third). In many cases, these could be reasonably accommodated through rehabilitation techonology rather than choosing the costly alternative of vocational rehabilitation.

Therefore, it is a challenge to those in rehabilitation who have the technical abilities to look at the field of Workers' Compensation as an area in which to apply their problem-solving skills. Clearly, the Workers' Compensation industry has a need. They can benefit from the use of a creative problem-solving approach in the solution of vocational rehabilitation problems.

PROVIDING A COST-EFFECTIVE SERVICE

The term Rehabilitation Engineering has referred primarily to the research process over the last few decades. Within the Workers' Compensation service delivery system, rehabilitation engineering is "A systematic application of human factors analysis and problem solving methodologies in order to improve the functional ability and vocational potential of the injured or disabled person." Given that definition, a rehabilitation engineer is an educated and experienced technologist who is knowledgeable in vocational rehabilitation and is trained to: understand the medical principles involved in disease and injury; apply engineering principles to the analysis of job task demands, human performance, and work place design; evaluate and/or design and develop the modifications necessary to increase the individual's functional ability or to reduce the job demands.

Workers' Compensation in California has several inequities that still need to be removed. One of them is that the system tends to be very subjective in nature. With this lack of objectivity, it becomes an easy process to manipulate. Consequently, it is the objective and systematic answering of the few basic questions which improves the process.

1. Do the job requirements exceed the medical limitations?
2. Will the job requirements aggravate the existing medical condition?
3. What specific job tasks are preventing return to work?
4. What, if anything, can be done to improve the working conditions and allow return-to-work?

In order to assist in objectively answering these questions, the rehabilitation technologist can provide several valuable services.

1. Human Factors Assessment - It is an objective and detailed functional job analysis based upon the ergonomic design of the work place, the occupational biomechanics of the job tasks, and the worker's physical abilities. This assessment identifies barriers which may be preventing effective return-to-work. In addition, it provides valuable information to a variety of people within the system.
    - a. To aid a physician in understanding the actual job tasks in relation to the client's medical limitations;
    - b. To aid an employer in the modifications or re-organization of a job task to meet the worker's abilities;
    - c. To aid the insurance carrier in determination of the claimant's qualified injured worker status (QIW) and available alternatives;
    - d. To aid a client in the development of a realistic vocational objective and rehabilitation plan;
    - e. To aid a vocational specialist by providing job placement information and job alternatives.
2. Job Modifications - Sometimes a worker could return to the original or similar job if the job tasks were slightly modified, if the job site was re-organized, if a tool was modified or developed, or if the client learned better work techniques. This can be provided directly or in conjunction with other professionals such as industrial engineers, supervisors, and physical and occupational therapists.
3. Ideals and Alternatives - By "brain storming" ideas based upon human performance levels, job task analysis, job site modifications, specialized tools and new technologies, many more employment alternatives can be explored. These ideas and alternatives are often generated during case staffings with other rehabilitation specialists such as company personnel officers and vocational counselors. Given recent legal findings, declaring more severe and difficult cases as "nonfeasible" may not be the best business alternative for Workers' Compensation insurance carriers.

PROBLEM SOLVING APPROACH

The Human Factors approach to vocational rehabilitation incorporates several interconnected disciplines into the work assessment process. This assessment process is frequently referred to as a Job Analysis and includes:

1. Occupational Biomechanics
    - a. Work postures required
    - b. Movement patterns
    - c. Work technique variations among workers
    - d. Physical condition of the employee(s)
    - e. Etc.
2. Ergonomic Factors in Work Place Design
    - a. Working surface
    - b. Work station design
    - c. Tools and equipment utilized
    - d. Methods or situations that increase the work demands
    - e. Etc.
3. Environmental Factors in the Work Place
    - a. Heat
    - b. Noise
    - c. Vibrations
    - d. Lighting
    - e. Etc.

Once the Job Analysis has been completed, then the job demands can be compared to the physical restrictions of the worker (medically determined). This comparison process identifies the barriers to returning an employee to his/her original job. At this point, the concept of Job Modification becomes a reality.

A Job Modification need not be expensive or require large-scale changes in the work environment. A job modification is any reasonable change to the work situation which accommodates the injured employee and allows return-to-work without placing increased hardship on other workers or the employer. These "reasonable changes" can be classified into four basic categories:

1. Adaptive behavior;
2. Work station modification;
3. Job task substitution or reassignment;
4. Use of assistive devices.

For example, a back-injured person involved in clerical tasks may benefit from one or more of the following job modifications:

1. Training in proper posture and body mechanics when sitting and bending - adaptive behavior.
2. Repositioning of the telephone to reduce reaching and twisting - work station modification.
3. Substitution of filing in low cabinets for increased photocopying - job task substitution.
4. Improved seating with an ergonomically designed chair and foot rest - assistive device.

CONCLUSION

A major question has arisen over the last several years as to whether rehabilitation engineering technology is cost-effective. In order to justify their activities and grants, institutions and rehabilitation engineering centers have created statistics regarding the long-term savings of rehabilitating the severely disabled. Vocational rehabilitation of the severely disabled is a truly worthwhile endeavor. However, it is not the only application of rehabilitation engineering technology and probably is not the most cost effective.

Rehabilitation of the severely disabled is only the tip of the iceberg in relation to the vast numbers of industrial workers who need services. In addition, the expense to this country related to Workers' Compensation is staggering - billions of dollars per year and thousands of capable workers.

Therefore, the challenge is to take a problem-solving approach which works with the relatively small numbers of severely disabled and apply it to the large numbers of mild to moderately disabled industrial workers. Granted, the nature of the successes generally are not as spectacular in the Workers' Compensation field. In fact, successes seldom, if ever, make the newspapers. However, the number of people whose lives are positively affected is great. For most people, the ability to work is synonymous with independent living.

The question of cost-effectiveness is answered in one word. SURVIVAL. In a fee-for-service industry, the economic value of the product or service and the ability to make reasonable market penetration determines the long-term result. There are no grants or subsidies in Workers' Compensation.

Fortunately, rehabilitation technology can and does function in the Workers' Compensation field. The approach outlined in this paper is one which has developed during ten years' experience as a private consultant in vocational rehabilitation. The assessment and problem-solving skills necessary to improve the function of the severely disabled are applicable and definitely needed in cost-effectively returning the industrially injured worker to his/her job.

16.6

PROSTHETIC TERMINAL DEVICE FOR PLAYING THE PIANO

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ABSTRACT- A terminal device is being developed to enable a person fitted with a below-elbow prosthesis to play the piano. The device, which consists of two fingers and a pivoting wrist, is activated by a foot control mechanism.

INTRODUCTION

The purpose of this project is to design a prosthetic terminal device for a 10 year old girl, R.M., who indicated a desire to play the piano. The child has a congenital deficiency of the left arm, a disability equivalent to a short below-elbow amputation [1]. She currently has a conventional hook prosthesis with a quick release wrist unit and a Figure-8 type harness.

Functional design criteria for a terminal device have been developed in conjunction with R.M.'s piano instuctor. The primary requirements are based on the accompanying role played by the left hand and include; 1) variable spread of the two "fingers" to provide combinations of a third to a full octave; and 2) wrist rotation for the black and white key combinations and full range on the lower half of the keyboard.

The project objectives are to design a terminal device so that it:

1) fulfills the motion requirements,
2) can be accurately controlled using cables,
3) provides positioning feedback,
4) is light weight,
5) can be built utilizing commercial parts.
6) is esthetically pleasing.

MATERIALS and METHODS

Task Analysis

R.M.'s music instructor specified five chords in a single octave that are often played with the left hand. As seen in Table 1, these include a third, combinations of a fifth, and a full octave. To quantify this range of motion a photographic hand study was performed using an accomplished female pianist. Eleven targets were placed on the left hand, wrist, and forearm and 35mm slides were taken at the prescribed chord positions. Additionally, distances for specified key combinations on the piano and are shown in Table 1.

CHORD	FINGER SPREAD (in)
Third	2.0
Diminished Fifth	3.625
Perfect Fifth	4.0
Augmented Fifth	4.375
Octave	6.5

Table 1.

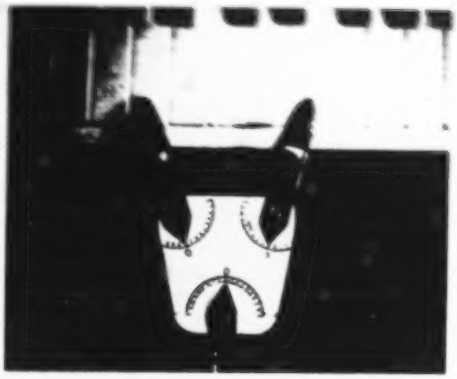


Figure 1

Prototype Design

A prototype terminal device has been developed for evaluation purposes and to allow R.M. to practice with a "two fingered" left hand. The prototype consists of an aluminum palm plate which has pivots at the fingers and wrist. Wing nuts are attached to the screws at each pivot to allow them to be positioned and locked in place. The design resembles a hand with features such as a contoured foam pad under the palm plate and flesh tone plastisole at the finger tips (which also provides protection to the keys). It is attached to the prosthesis using a Hosmer FM 100 [2] quick release attachment.

Client acceptance of the prototype has been exceptional. She has affectionately given it the name Bach and uses it on a daily basis for practice.

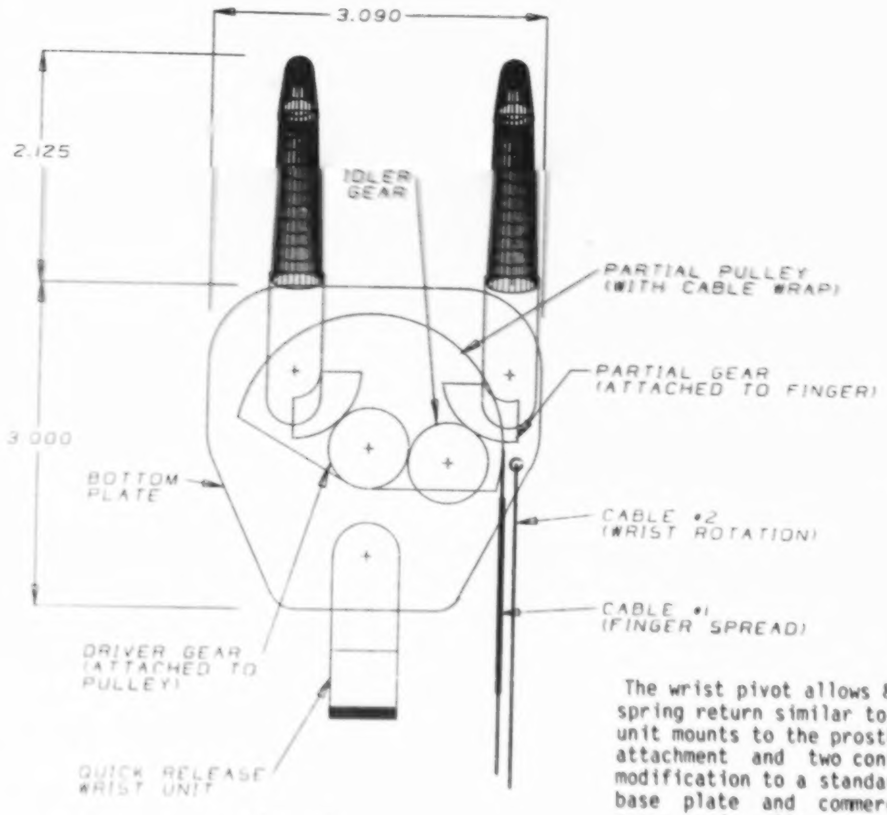
Evaluation

Pointers and scales were placed at the pivots of the prototype (Figure 1) to quantify motion of finger and wrist. Maximum wrist rotation needed with the prototype (80 degrees) was found to be substantially larger than that in the photographic study. The reason for this was that the fingers on the prototype lacked the adaptability and complex motions of the human fingers. Maximum values of wrist rotation for the photo subject and prototype are shown in Table 2.

	Max. Wrist Rotation
Subject	+17.5/-8.0
Prototype	+37.0/-46.0

Table 2.

Shoulder protraction, normally used to open and close R.M.'s hook prosthesis, was initially considered for control purposes. It was rejected however, due to the difficulties it imposed on R.M.'s ability for keyboard positioning. Foot control


3.090
2.125
3.000
IDLER GEAR
PARTIAL PULLEY (WITH CABLE WRAP)
PARTIAL GEAR (ATTACHED TO FINGER)
BOTTOM PLATE
CABLE #2 (WRIST ROTATION)
CABLE #1 (FINGER SPREAD)
DRIVER GEAR (ATTACHED TO PULLEY)
QUICK RELEASE WRIST UNIT

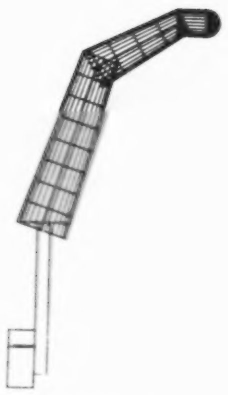

"FINGER" PROFILE

Figure 2

was determined to be the best available means for controlling finger spread and wrist rotation. It easily accommodates two degrees of freedom for these 2 motions and does not interfere with upper extremity control.

FINAL DESIGN DESCRIPTION

Methodology

Following a design methodology similar to that developed by Sears [3], the design and redesign phases have been accomplished using a computer aided design system. The entire device consists of two seperate units, the terminal device and a control mechanism. The control mechanism is foot operated and provides two independent motions which are transmitted by cable to the terminal device.

Hand Mechanism

The device (Figure 2) efficiently fulfills the motion requirements. The fingers rotate in opposite directions and range from a third to a full octave using only one cable. A rotational spring under the driver gear provides the return force. The fingers are rubber tipped to provide a soft, "sticky" interface with the keys. They are contoured from aluminum tubing in a shape found functional and cosmetically attractive from prototype evaluation.

The wrist pivot allows 80 degrees rotation with a spring return similar to the fingers. The entire unit mounts to the prosthesis with a quick release attachment and two control cables requiring no modification to a standard prosthesis. An aluminum base plate and commercially available plastic gears and pulley are used to maintain a light weight device. The total weight is estimated between 12 and 16 ounces. A cover molded from kydex will encase the hand mechanism.

Foot Mechanism

The foot control (Figure 3) consists of a small box placed under the left foot of the pianist. Sliding the pedal the length of the box produces a finger spread of a third to a full octave. A cam plate has been designed so that spacing for the five combinations is equal.

The pedal attaches to an aluminum carriage which is guided by parallel shafts (Figure 4). Linear travel of the carriage produces lateral cam plate motion for cable extension and finger spread.

Thin vertical grooves ("frets") are detected by a bearing plunger as the pedal is moved linearly for finger control. This provides position feedback to the user. Bumper stops are placed at the ends to dampen impact and reduce noise in the device.

Rotation of the pedal about its center directly relates to the amount and direction of wrist rotation. The amount of wrist rotation chosen by the pianist is independent of finger spread, permitting chord combination anywhere on the lower half of the keyboard.

Learning features in the form of locking mechanisms to prevent either the translational or rotational motion have been provided. These can be removed once R.M. has mastered the device.

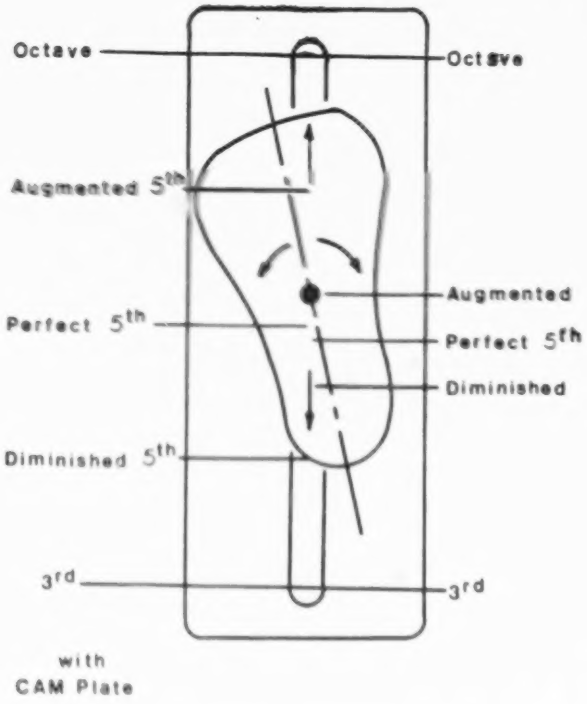

Octave
Octave
Augmented 5th
Augmented
Perfect 5th
Perfect 5th
Diminished
Diminished 5th
3rd
3rd
with
CAM Plate

Figure 3

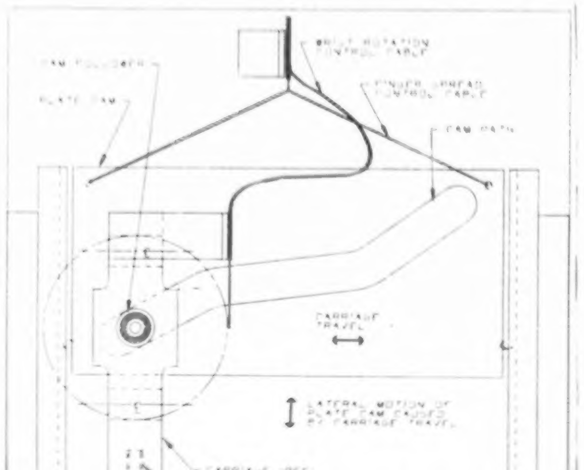

WRIST ROTATION CONTROL CABLE
CAM FOLLOWER
PLATE CAM
FINGER SPREAD CONTROL CABLE
CAM PATH
CARRIAGE TRAVEL
LATERAL MOTION OF PLATE CAM CAUSED BY CARRIAGE TRAVEL
CARRIAGE (REF)
PLUNGER (REF)
LINEAR BEARING

Figure 4

DISCUSSION

The terminal device and foot mechanism are designed to be functional for playing the piano and cosmetically pleasing. The light weight device provides feedback and learning features for the user. The device mounts directly to R.M.'s prosthetic arm with no modifications or adjustments required.

Efforts are now directed towards utilizing the CAD system to provide kinematic and structural information for the design. A list of commercially manufactured parts is also being assembled. Fabrication of the final design will be completed to keep pace with R.M.'s developing musical skills.

ACKNOWLEDGEMENTS

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An Improved Optical Headpointer Selection Technique

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ABSTRACT

An improved optical headpointer selection technique has been developed for use in expressive communication and computer access systems. The technique offers faster selection speed for some existing users and allows operation by others who would previously have had to use much slower scanning techniques.

INTRODUCTION

For nearly everyone, effectiveness in communication determines to a large extent where a person goes in life and what can be accomplished. For people with severe communication impairment, communication effectiveness generally translates directly to communication speed. Communication speed is a composite of three factors:

1. Selection Speed
2. Number of Selections Needed
3. Amount of Information Retrieved.

This paper deals only with the first of these.

There are many selection techniques used by people with severe physical handicaps in the operation of expressive communication aids and computer access systems. Selection speed is very much a factor of the selection technique being used. Prior to 1979, many people without keyboard skills were provided with scanning systems operated using a single switch. Although dreadfully slow, these systems did offer somewhat more than nothing at all. While small percent speed improvements can be realized through optimizing character arrangement, control interface, etc., the greatest speed improvement can generally be realized by using a direct selection (pointing) technique. Unfortunately, many of these people had physical skills to permit the use of a direct selection technique, but no appropriate hardware existed.

Then in early 1978 engineers at the University of Tennessee Rehabilitation Engineering Center in Memphis developed the optical headpointer for a local school student. Prentke Romich Company made the system commercially available the following year and it has been employed in the Express 1, Express 3 and Minspeak 1.

The optical headpointer has been a useful selection technique for many people who would otherwise have had to use the much slower scanning systems. The technique has also allowed the design of systems such as the Expresses and Minspeak 1 in which one device offers a wide range of selection techniques, including optical headpointing, scanning and others. This can be important for people with changing physical capabilities or those with different abilities when in bed relative to those when sitting in a wheelchair.

OLD SYSTEM

The system as originally implemented consists of an array of red light emitting diodes (LED's). Each LED corresponds to a letter, word, symbol or other selection on the communication aid. The user points at the array using a light sensor, perhaps combined with an optical system (to permit operation at a distance) and mounted on the head. The electronic circuitry operates in such a way that the LED toward which the sensor is directed is bright. This, then, is the feedback showing the user where the light sensor is directed.

The user operates the system by directing the light sensor toward the desired location. To enable the user to move from one position to another without selecting all positions between the two, a delay has been incorporated. To make a selection, the desired LED must be kept on for a period of time.

However, many people lack the necessary head control to use the optical headpointer as it had been implemented. The difficulty arises from the fact that the task is to maintain a particular location for a continuous period of time while the only feedback the user receives is that the location is being indicated. A number of variations were suggested and implemented to improve the effectiveness of the system. The Trace center suggested the use of a projected light beam aligned with the light sensor optics. This provided a second source of feedback for the user and indeed improved performance. Another approach was to enlarge the target by treating a group of four or a group of sixteen LED's together. These ideas were implemented in the Express 3 and Minspeak 1. Each of these variations represented a compromise in some area.

A similar problem was addressed in the development of the Autocom over ten years ago. With the Autocom, the user would make selections by sliding a magnet across an array of magnetic sensors (1). Although the Autocom patent claimed coverage of systems allowing non-continuous activation of the sensor, the Autocom was implemented requiring continuous activaction.

NEW SYSTEM

The improvement came out of observing a person with amyotropic lateral sclerosis (ALS) using the original implementation. The user had head position control that made optical headpointing much faster than scanning. Still, his skill and speed were such that it was somewhat easier for the human observer to determine the desired target. It was suspected that the difference was that the observer retained a history of the pointer position while the system did not.

The improvement that has been implemented does not require that the LED be indicated for a continuous time. The old system used one timer that was reset each time a new LED was indicated. The new system uses 128 timers, one for each LED. When any timer expires, that location is accepted and all timers are reset. This allows the system to "remember" where the optical headpointer has been directed since the last selection was accepted. The improvement is now employed in PRC devices using optical headpointing. Most Express 3 and Minspeak 1 units already in the field can be updated through a simple exchange of program memory.

In addition to the optical headpointer, the technique could be used with any direct selection system using visual targets with a cursor and a means of moving the cursor. This could include systems based on various display technologies and input systems such as the conventional joystick, ultrasonic head position sensing (2), etc.

The improvement appears to be a far more powerful technique for people with some head instability. It has greatest significance for those people who would otherwise need to use a scanning selection technique.

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PNEUMATIC PRESSURE SWITCH FOR COMMUNICATION

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ABSTRACT

The project reported here enabled a severely physically limited person to access and operate a computer through switch closures. The intermediate goal was to identify an optimum switch which matched the available motions. Available motions were identified as eye blink and thumb flexion. There were no predeterminations to "make" or "buy" the device. A team approach was selected for implementation.

INTRODUCTION

Identifying the proper assistive device can become a challenging task. The selection or identification process is not a unique endeavor but can be time consuming if the proper techniques are not employed. A frequent technique employed is the literature search of catalogs. Even though devices can be successfully identified and selected from catalogs, they may not exactly meet the disabled person's needs. Quite frequently those items selected from the literatures need further modification before they can be used. Likewise, devices made to custom fit the disabled person's needs can be equally inappropriate. The techniques employed in everyday shopping, mainly "try" before you "buy", do not exist for the handicapped. There are very few stores or outlets which stock a wide selection of devices where a disabled person can try the item before it is purchased.

A combination of techniques were used to identify a workable device for the disabled person cited in this paper. The work accomplished was the result of the effort of a team whose composition included medical clinicians, attendant nurses, designers and an information specialist. Limited successes have been achieved but more investigations and evaluations of devices are yet to come.

BACKGROUND

The person who will be cited here has a disability resulting from a stroke. A lesion to the brain stem has resulted in a disabling condition known as locked-in-syndrome. She is a nonvocal quadriplegic who communicates using eye blinks. Even though her eye movements are poorly controlled, she uses coded eye blinks successfully. Her intelligence remains intact. She uses a computer as a means of written communication. She has a small amount of movement at the IP joint, left thumb. Maximum movement is approximately four millimeters flexion with minimal force being detected.

Switch Selection

A switch closure is necessary to access and operate the computer. The computer is coupled with the Prentke Romich Company Express I system. At the outset of this project, an eye blink switch could not be identified. Subsequent to the initial search for blink switches, a switch has been identified and purchased for evaluation. The unavailability of a suitable blink switch necessitated the thumb switch development. A number of thumb-type switches were developed and evaluated. The first series of switches were deemed as not being appropriate or required too great a force for her to activate. As shown in figure 1., a switch was created using a low force microswitch. This microswitch device could be activated, however, the mechanical travel required to obtain a switch closure was at the maximum limit of her thumb flexion. A pneumatic pressure switch which could be activated using a pressure less than 1/2 inch of a water column was investigated. The trial proved to be successful and a switch closure device was fabricated for trial and evaluation.

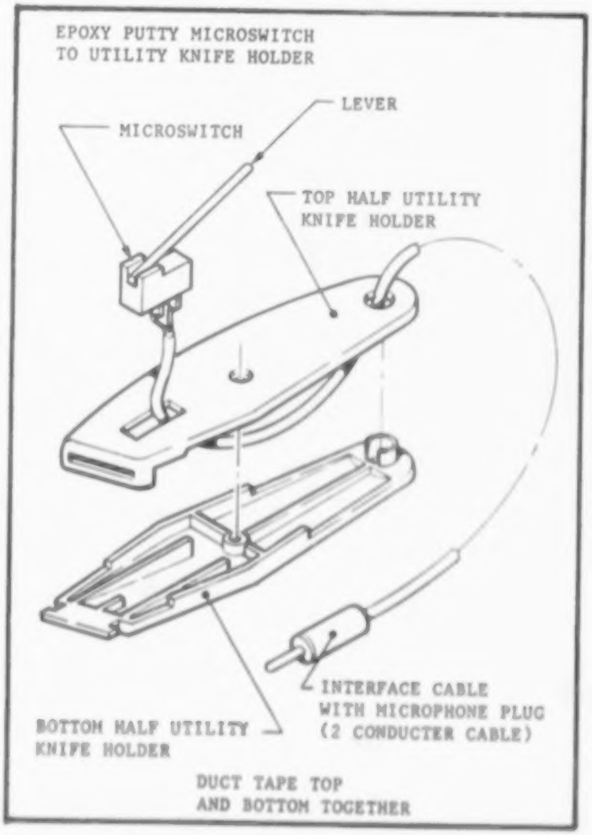

EPOXY PUTTY MICROSWITCH
TO UTILITY KNIFE HOLDER
LEVER
MICROSWITCH
TOP HALF UTILITY KNIFE HOLDER
INTERFACE CABLE WITH MICROPHONE PLUG (2 CONDUCTER CABLE)
BOTTOM HALF UTILITY KNIFE HOLDER
DUCT TAPE TOP AND BOTTOM TOGETHER

Figure 1. Microswitch Device

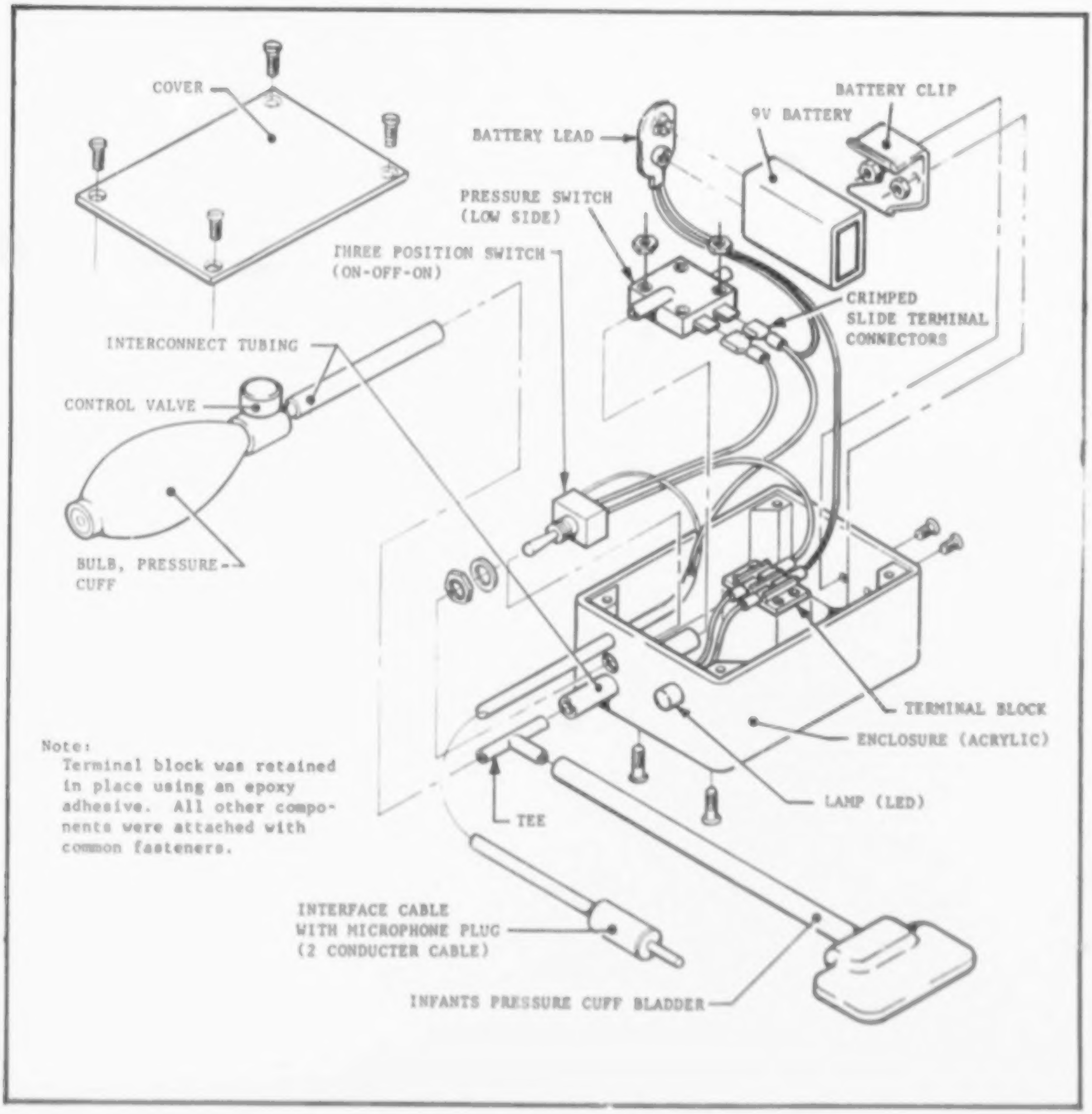
MATERIALS AND METHODS

First Generation Pneumatic Switch Device

The first pneumatic switch was a primitive device designed to be used for assessment. Components were taken from previously discarded pieces. The composition of this device was a LED lamp, 9 volt battery, pneumatic pressure switch, infant blood pressure cuff bladder, electric wire and duct tape. The purpose was to provide a practice device which could assess travel and force available from the thumb movement. It was found to be effective for this person and the LED lamp gave her feedback information. After two weeks of intermittent practice, the preliminary device was deemed appropriate and switch closures could be accomplished on a repetitive basis. At first, activating the pneumatic switch was fatiguing to the user. Through practice and daily use, her fatigue factor diminished somewhat.

Second Generation Pneumatic Switch Device

The second generation device was fabricated incorporating some suggestions of attendants and the user (see figure 2.). Considerations used on the follow-on device were enclosure


COVER
BATTERY LEAD
9V BATTERY
BATTERY CLIP
PRESSURE SWITCH
(LOW SIDE)
THREE POSITION SWITCH
(ON-OFF-ON)
CRIMPED
SLIDE TERMINAL
CONNECTORS
INTERCONNECT TUBING
CONTROL VALVE
BULB, PRESSURE-
CUFF
TERMINAL BLOCK
ENCLOSURE (ACRYLIC)
LAMP (LED)
TEE
INTERFACE CABLE
WITH MICROPHONE PLUG
(2 CONDUCTER CABLE)
INFANTS PRESSURE CUFF BLADDER

Note:
Terminal block was retained in place using an epoxy adhesive. All other components were attached with common fasteners.

Figure 2. Pneumatic Pressure Switch Device

design, pressure cuff inflation bulb, and a three-position toggle switch for selection of operating modes. The enclosure was necessary to protect the electronics components and battery while the addition of a pressure bulb was needed to maintain the proper inflation pressure in the cuff bladder. The three-position switch was incorporated to provide a center "off" position and the remaining two positions controlled the LED light feedback mode or straight switch closure without feedback. It was also discovered that during periods of spasms, the relationship of the thumb to bladder would be lost and repositioning needed to be accomplished. To overcome frequent repositionings by the attendant, a closed cell urethane block was fabricated to capture and stabilize the hand. When the hand is placed in the fixture, a velcro enclosure secures the hand. This allowed the thumb and bladder to remain in contact.

Third Generation Pneumatic Switch Device

A final generation of the switch device is currently being proposed. This unit will integrate electronics, pressure switch and hand stabilization features into a unit. Other considerations are to provide a dual switching arrangement by integrating pressure and blink switch functions into the single unit. The primary concern has been computer access, however, coupling to other devices will be given equal priority. The construction will include molding plastics and state of the art component packaging. As before, those who have participated in formulating the previous design will continue to input their expertise.

RESULTS

Current experiences using the pneumatic pressure switch indicates satisfactory performance. Concurrent with the pneumatic pressure switch trials, an eyebrow switch and bite switch were tried for alternatives. The eyebrow switch was not suitable because there was insufficient movement of the eyebrow. The bite switch was also unsuitable because biting initiated frequent spasms. The recently acquired blink switch offers the best alternative for the pneumatic pressure switch. A comparative evaluation of these two switches will be conducted at a later date.

CONCLUSION

In cases where slight finger movement is available, a pneumatic pressure switch can provide alternatives for switch closures. As previously indicated, identification of switching devices is often a challenging event. Although the literature provides information, there still remains the need to "try" before the final purchase is made. There are evaluation devices available to aid in switch assessments, but many communities lack the resources to acquire these devices. There are occasions when a "make" decision is rendered because information resources are lacking. The "make" decision, most generally, risks reinventing the wheel. Likewise, decisions to "buy" from literatures risk inappropriate selections. Are there alternatives? Ideally it is better to "try" before you "buy". If "try" is not possible, then "buy" and "modify" (adapt). If your alternatives have been exhausted, reinvent the wheel. We did.

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17.3

FACTORS IN THE DESIGN AND DEVELOPMENT OF AN INTERACTIVE HUMAN-ROBOT WORKSTATION

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ABSTRACT

The recently funded proposal for an advanced manipulation aid for severely disabled individuals mandates improvements of the user interface to the robotic system. This paper describes the design and development of a workstation, the hardware portion of the human-robot interface, for the next generation Veterans Administration/Stanford University Robotic Aid. The workstation is designed to provide maximum access to the greatest number of users, both those with disabilities and the able-bodied; to meet the needs of the technical development and clinical evaluation teams; and to allow for multi-user interaction and easy group demonstration.

INTRODUCTION

The VA/SU robotic aid is presently undergoing second generation development. The VA Merit Review Grant[1] through which the project is funded mandates design and development of the user interface where key specifications are based on the experiences of, and needs expressed by, the users of the first generation system. This interface must accommodate multiple input and output (I/O) channels. Various devices, such as ultrasonic head position sensors and a six degree-of-freedom joystick, will be integrated into the next generation design.

An important subsystem of the human-robot interface is the user workstation; it is the physical component that links the user to the machine. Levy[2] defines an interface as "a means of access for an individual to an object, mechanism, machine, or system of machines the individual wishes to exploit optimally in the process of carrying out a specific activity" and considers the interface of primary importance in matching individuals to adaptive devices. He describes a workstation as a necessary peripheral for maximizing effective use of a technical aid. The specifications for Levy's workstation include: ergonomic factors, user functional requirements, accessibility, aesthetics, safety, materials, integration into the user's environment and connection to other systems.[3]

In the design of the interface to the robotic aid, the system designers must strive to maintain the relationship between the human and the robot while addressing human factors issues such as anthropometrics, task-related human factors and value-related human factors. The designers must also consider two means of interaction — a general system console that effectively integrates the various input and output devices necessary to monitor and control the robotic aid and its peripherals, and an interface to the mobile base on which the robotic arm will be mounted. The mobile unit will require a miniaturized console to debug software and to display system status information.

BACKGROUND AND WORK ACCOMPLISHED

The first generation robotic aid served as proof of the concept that microprocessor controlled human-scale industrial robots can be used by persons with physical disabilities in completing manipulation tasks. User-specific training has focused on activities of daily living, recreational and vocational tasks such as preparing a meal, playing board games and opening a file drawer to retrieve a file. The objective of second generation development is to demonstrate the utility of a human-service robotic aid and to deliver the technology needed to make robotic manipulation aids broadly useful.

Evaluation and testing of the robotic aid with disabled users has been performed according to the Interactive Evaluation Model developed by Engelhardt, et al.[4] Over 90 people, 23 of whom are spinal cord injured, have been trained to use the first generation system in the past two years. Some of the comments users have made are that the robot is too slow, it requires too much attention and is too dependent on pre-structured environments. User feedback has been essential both in guiding modifications of the first generation aid and in setting the key specifications for the advanced manipulation aid.

The clinical system employs voice as the primary I/O channel, with alternate input channels, the Ultrasonic Head Control Unit (UHCU)[5] and VIDOF, a six degree-of-freedom manipulandum[6] for people with residual motor function. The UHCU has been used to pilot the robotic arm and VIDOF to give the user simultaneous control of all axes of a six degree-of-freedom industrial arm. Other input devices include a teachbox and a standard keyboard for use by the system operator in programming motion sequences, and a pushbutton for emergency stops. An 8-line, 160-character plasma display is used, along with synthesized speech, for output to the user. A DEC VT-100 terminal displays diagnostic information (e.g., the last command recognized, the time elapsed in the current session, and the command recognition confidence level) to the system operator. The sensory capability of the clinical system is limited to a few touch activated microswitches in the gripper of the arm.

The second generation lab system employs two LSI-11 computers that directly command the robot arm joint controller. It also includes improved speech synthesis and recognition units, VIDOF, the UHCU, a series of programmable switches, a keyboard and a television camera. VIDOF is used for six-axis control of the robotic aid while the UHCU is dedicated to two-axis control of the mobile base and the graphics screen cursor. The programmable switches will be dedicated according to need. Current functions include arm power control and an emergency stop button. The television camera mounted on the robot is the input portion of the vision system that will provide real-time information to the user on the working environment of the robotic aid. The mobile base will house one of the LSI-11 computers and will communicate with the central processing unit through a radio link. A high resolution graphics display will be used, in addition to a diagnostic display and a video monitor, to enhance the presentation of system states to the user (see Figure 1). Examples of the kind of information that will be displayed on the graphics terminal are a menu of commands available to the user and graphic environmental data to aid the user in trajectory planning. The Tektronix 4107 and the Macintosh are the present candidates for the high resolution graphics system.

The sensing capabilities of the robotic aid will be greatly increased in the second generation, lessening the control burden on the user. In the advanced robotic aid, the user and the robot will work together to complete a task. The user, in a supervisory role, will need information about the environment that can be provided by various sensors especially when visual information

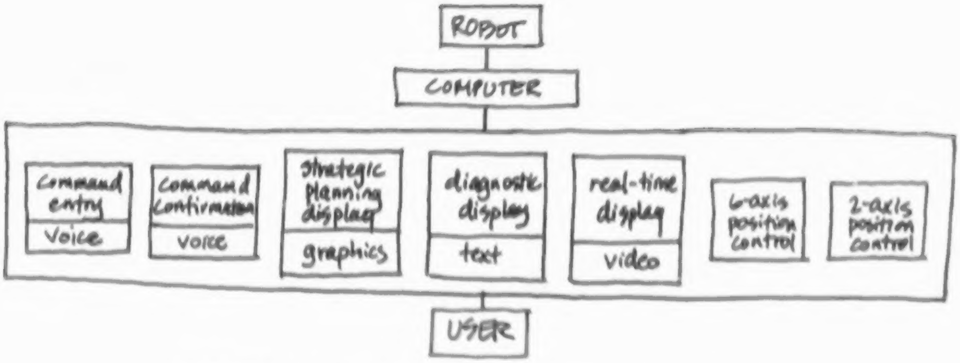

FIGURE 1. Functional elements of the human-robot workstation.

is not sufficient. This system is to include: a range sensing hand with obstacle avoidance, automatic grasping and object modeling capabilities; a mobile robot with obstacle sensing and remote docking capability; and a force-sensing wrist with active force control and force thresholding features. In this manner, sensors can increase the utility of the robotic aid as new tasks can be attempted.

DESIGN CONSIDERATIONS

The human-robot workstation must be highly flexible to allow for changes in focus, user and environment. It must be reconfigurable to meet the varied specifications of prototypic situations. The six most common situations are listed: (1) Close personal interaction between the human and the robot, in tasks like getting a drink of water, makes safety a vital issue. (2) Local interaction, object identification and placement (i.e., to set a dinner table), require that thought be given to access and workspace for both the user and the robot. (3) Interaction that involves the mobile base necessitates a more global world view and a method of monitoring and controlling the gross motion of the base and the fine motion of the arm. (4) In development of the system, different people may be working on different components of the system simultaneously. (5) During a user training session, the trainer may need access to system software while the user interacts with the robotic aid. (6) In demonstration, more than one person may be viewing the displays at a time (see Figure 2).

The design of the workstation must allow for the characteristics of different user populations — quadriplegics and others with reduced motor skills, limited range of motion, and reduced strength; the elderly and others with limited dexterity and low vision; children with their different level of understanding and methods of learning; and the development and evaluation teams. In addition, the workstation must be [illegible]ble, so that the system can be tested outside of the lab[illegible]ory setting, accessible to all users, approachable and easy to maintain.

METHODS

The first step in the design of this workstation was a literature and product search in the area of interactive human-machine systems. The topics in the literature search ranged from telemanipulation and space station design to shared attention environments and display devices. The product search included investigation into flat panel display technology, touch screens, "knee-top" computers and keyboards with infrared links. Device evaluation centered on power specifications, compatibility with existing equipment, human factors (i.e., the viewing angle and contrast ratio of a particular display), size, weight and, of course, cost.

Location and arrangement of visual displays and controls were determined by consulting general human engineering guidelines for designing individual workplaces (see Figure 3). Each of the three displays, the color graphics, diagnostic and video displays,

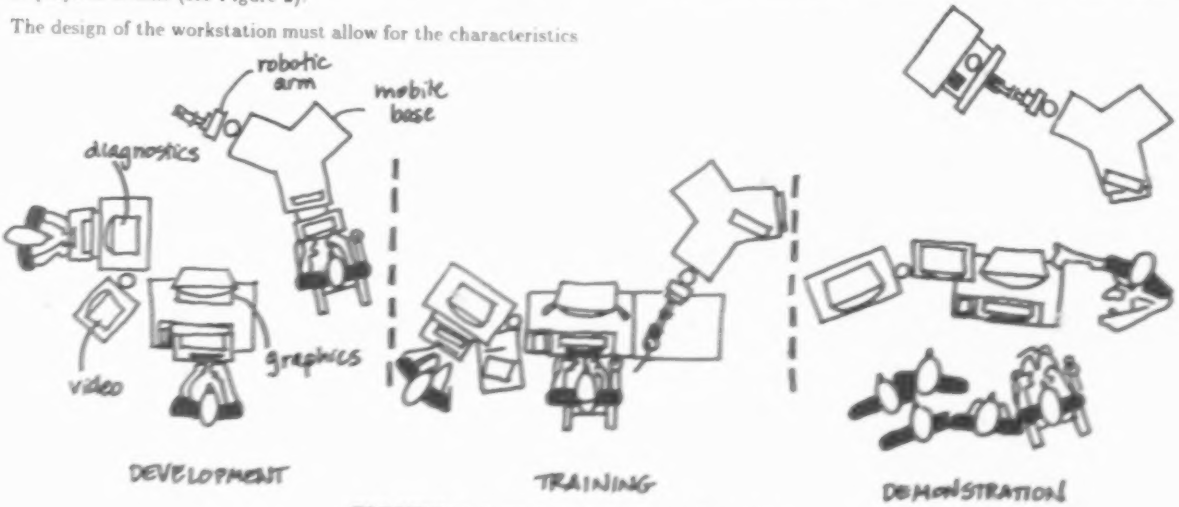

FIGURE 2. Sample workstation configurations.

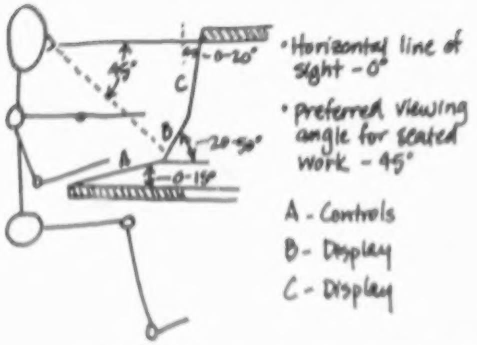
0-20°
45°
C
B
A
20-50°
0-15°
• Horizontal line of sight - 0°
• preferred viewing angle for seated work - 45°
A - Controls
B - Display
C - Display

FIGURE 3. Dimensions for visual work.

was then given a certain rating based on importance, frequency of use, function, and sequence of use. The color graphics display achieved the highest rating as it is necessary for monitoring robot motion. It is also needed in both development and performing detailed work. The diagnostic display also rates as a primary visual display in development as does the video monitor in detailed work applications, so they too must be accessible.

The design of this workstation has been likened to the design of a vehicle dashboard. Both require that special attention be given: to the forward visual field of the driver or operator while maintaining the visual field of display; to hand (or mouthstick) reach for control action; and to knee and leg (or wheelchair) clearance for accessibility and comfort. An effort was made, in the acquisition of anthropometric data, to address the physical needs of the greatest number of users while considering their visual and psychomotor capabilities.[7,8]

RESULTS

One version of a human-robot workstation is shown in Figure 4. It consists of a main console unit through which the robotic arm and mobile base are accessed, and a reconfigurable "tree" that changes with the situation. The architecture permits single or multi-user interaction with the robotic aid and makes possible simultaneous viewing of the displays. In addition, the color graphics terminal rotates from 0-50 degrees off the vertical to keep the visual line of sight clear when the mobile base is in action. The keyboards for both diagnostic and color graphics displays are designed to be accessible. The pedestals on which they sit translate up and down and the keyboards can easily be moved to a lap tray or to another preferred work surface. Workspace is available to both the user and the robot in all applications.

The mobile base mini-console (not shown) would consist of a keyboard with programmable keys and a flat panel display, preferably an electroluminescent one due to its graphic capability, wide viewing angle, low power consumption and high resolution. Gas plasma technology is comparable but the displays are difficult to acquire at this time; liquid crystal displays, while inexpensive, have an extremely limited viewing angle and contrast ratio and are, therefore, inappropriate.

FUTURE PLANS

Iteration and refinement of design concepts are expected. The initial prototype, under construction now, will be used with the laboratory system. Following system debugging and further development, the updated robotic system, mobile base and workstation will be transferred to the clinical group for evaluation and user trials. This cycle of development and evaluation will continue until the delivery of what will hopefully be deemed a practical human service robotic device with an appropriate human-machine interface.

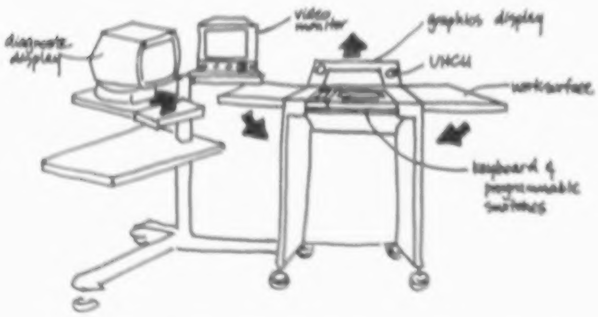

FIGURE 4. Designer's concept of a human-robot workstation.

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INTERACTIVE EVALUATION OF ARTIFICIAL VISION WITH ROBOTIC AIDS FOR INDIVIDUALS WITH DISABILITIES

17.4

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ABSTRACT

The Interactive Evaluation Model[1] has been applied with the assistance of four users, to the use of artificial vision for a robotic aid. Evaluation has been developed on an industrial vision system to determine how existing vision technology might be utilized in rehabilitation applications. Computer program compatibility, item identification and location, and adjustment to ambient lighting are crucial technical issues that have already been identified. We propose that the use of coded markers on daily living items and on the surroundings will greatly alleviate challenges in these technical areas. Use of Artificial vision with robotic devices in semi-structured and home surroundings may reduce the financial costs associated with a "seeing Robotic aid".

INTRODUCTION

The feasibility of an interactive robotic aid for disabled has been studied using the Interactive Evaluation Model.[1] This model emphasizes the need for a steady flow of feedback from potential users to designers and developers during the entire lifecycle of research, development, and evaluation of advanced assistive devices. Useful voice control of a robot arm has been demonstrated earlier in a structured environment.[2] While feasibility of voice control has been demonstrated, the user experiences a cognitive workload that can lead to longterm fatigue in disabled and able-bodied uses. There is now the need to evaluate more autonomous robotic aids in a less structured enviornment, possibly remote from the disabled user, which will give the user a wider range of real functional control. Artificial vision [3] offer that possibility. This paper undertakes an interactive evaluation of a particular artificial vision system that will be incorporated into a robotic aid which is voice controlled by disabled users.

BACKGROUND

The Palo Alto Veterans Administration / Stanford University robotic aid has employed voice control of a Unimate arm and evaluated the feasibility of using this device for persons with disabilities. The user visually asseses useful tasks the arm might perform in daily living settings directly before him or her, and voice directs details of the task completion by the arm. The user may, for example, guide the arm, when it is in sight, to pick up a glass of water. A more desirable eventual goal might have the user state, "Give me the glass," and then have the robotic aid identify, locate, grasp and present the glass to the user. Present artificial vision systems offer a first step to accomplishing such an eventual goal.

VISION SYSTEMS

Many artificial or machine vision systems have computational algorithms based on the Stanford Research Institute algorithm.[4] An adjustable illumination threshold is set in the system and a gray scale electronic picture of a scene is then converted to BLACK or WHITE. Pixel signals below threshold are changed to black and signals above threshold set to white. To reduce system noise, a minimum area can be set below which the system will not recognize signal cluster as an item. As a consequence of the threshold and noise settings, the ambient lighting must be carefully adjusted during normal operations. The system *extracts features* from the black and white scene including:

- Number of items, both objects and holes within the objects
- Cordinates of items relative to scene dimensions
- Relative item areas
- Relative perimeters sizes
- Orientations
- Largest and smallest dimensions

The pick-up camera can be remote from the user who can view a video presentation of the scene. Scene processing rates may be 5 to 10 per second. Newer generation vision systems are faster and can control more cameras but do similar processing.[5] In terms of human vision, artificial vision systems are excellent at locating and giving relative sizes of items but very poor at identifying random items especially in an unstructured environment.The relative geometrical information extracted by the vision system on an item that has been identified must be converted to actual spatial dimensions. Then the vision system can inform the user or robot where the item is located and how to grasp the item.The voice commands which will operate the vision system must fit comfortably into the voice command syntax and other computer software of the entire robotic aid. Crucial technical issues in the use of a vision system with a robotic aid are thus:

- Identification of objects
- Determination of actual distances
- Operation in ambient lighting conditions
- Software integration of the vision system Program compatibility of the vision system with the robotic aid.

METHODOLOGY

We are employing a GE Industrial Vision System [6] and have designed task simulations that are relevant to disabled users. The GE Vision System will extract all of the vision features noted, and others, but our version of the system has limited memory and is fixed in location. Four wheelchair users have been included in the project from the beginning. One wheelchair user is 27 years old and has had multiple sclerosis for 10 years. One user is quadriplegic and the other two are paraplegic. The

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Interactive Evaluation Model as applied to the use of artificial vision with a robotic aid is illustrated below.

CONTINUED RESEARCH ON ARTIFICIAL VISION FOR INTERACTIVE ROBOT AID

- optimum coding
- software redesign
- user and machine profiles
- training procedures
- user feedback

TECHNICAL DEVELOPMENT

Machine Performance

- query user
- accept vocal instructions
- present visual scene
- identify coded and uncoded objects under user supervision
- determine spatial data
- instruct manipulator
- interact with user
- complete task
- collect evaluation data
- allow modular programming

Desireable User Characteristics

- speaks with volume and regularity acceptable to voice system
- ability to read screen display
- ability to understand information displayed

CLINICAL TESTS

Human Performance

- vocal consistency, strength and quality
- eye control
- cognitive abilities
- workload limits

Desireable Machine Characteristics

- voice actuated
- accept continuous voice input
- adjust to lighting conditions
- identify objects
- determine distances
- store and remember object data
- accept program changes
- read codes and markers
- scanning capability
- presents information in understandable form
- integrated sensor set
- integrated software

TECHNOLOGY TRANSFER will require transfering the substantial industrial experience with vision systems to applications in semistructured and unstructrured home and health environments. The results of this research will also have implications for space and for a wide spectrum of industrial applications.

ISSUES

Modular programming[7] of the computer software will facilitate the incorporation of the vision system into the robotic aid software. One main CPU will control the artificial vision system, the robotic aid, the voice system and the data bases required for voice recognition, item identification, item characteristics, path verification and system evaluation.

In our evaluation tests, the problems of (a) identifying items, (b) determinging exact distance, and (c) adjusting to ambient illumination, have greatly limited the autonomy of the vision portion of the robotic aid. We have substantially alleviated these difficulties by the use of machine readable codes and markers. A coded enviornment offers a simple way to greatly reduce the financial cost of the robotic system with vision.

Use of codes and markers is quite compatible with robotic aids for disabled. The surroundings of a disabled person who might use the robotic aid are often constrained to a modified home or single room. It is feasible to mark items and the elements of the surroundings with codes which simultaneously announce to the vision system the name of objects, present calibration distances and allow adjustment of the system settings to variable illumination. Coding can be accomplished when the items are first provided to the user.

In general, the design and implications of codes should be robust and this implies several criteria. The codes should be

- redundant in character
- applied in multiple to objects, if necessary
- based on geometrical factors which are invariant to camera angle and viewing aspect.

The codes should also be flexible and pleasant in appearance so that they are readily assimitated into the user's surroundings. The codes are not limited to movable items but may be affixed to the surroundings. It would only be necessary to mark a very limited number of [illegible] such [illegible] doorways and fixed cabinets and appliances to allow a moving robot which has a floorplan encoded in a relatively small memory to move about a home or institution. In an interactive robotic aid code, failures are remedied by a system request for information directly to the user.

Our first phase of evaluation will identify principle in various coding schemes that will improve the design and implications of coding mechanisms. We have provisionally adopted a modified 2 of 5 coding and marking scheme for use in our testing.[8] Figure one illustrates the code. Filled symbols, such as dark circles, are code zeros and will be recognized by the vision system as items. Symbols with light centers are code ones and will be recognized as holes. The largest symbol is at the start of the code which must contain two zeros and have a fixed distance between the last four symbols. This fixed dimension serves as a distance marker, while illumination threshold is adjusted by program so that at least two holes appear in each coded object. Bar coding also has obvious utility since it is available on numerous items and only requires commercially available equipment and software.

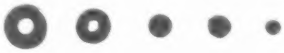

Figure 1. Code for use on a light colored item. The 5 symbol code reads from the largest symbol as 00111. The code must contain 2 zeros as a self check and so a single code group can distinguish 10 items.

SUMMARY

We have considered the feasibility of a vision-assisted robotic aid for disabled individuals. Present vision hardware appears capable of useful integration into the robotic aid if the environment is coded. The use of codes makes it feasible for vision systems to be brought out of the very structured industrial environment into the semistructured and unstructured environment found in rooms, institutions and the home to assist elderly and disabled individuals. New training procedures will have to be developed in order to facilitate acceptance by disabled individuals of a fully integrated robotic aid with vision.

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17.5

TOWARDS VOICE CONTROL OF ROBOTIC AIDS FOR DISABLED

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ABSTRACT

The increase in popularity of micro computers and low cost robots for home, education and light industrial use has opened up the possibility of a general purpose and affordable aid for physically disabled.

The suitability of a low cost voice recognition system for such devices is evaluated with special emphasis on its use by cerebral palsy and muscular distrophy children.

INTRODUCTION

For a robotic aid to succeed it must be seen to be safe, cheap, easy to operate at a simple level and interface simply with the form of communication selected by the therapist or user.

HARDWARE ARRANGEMENT

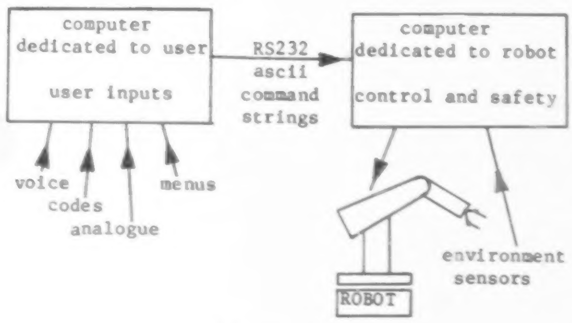

figure 1.

Since such an aid should be available to as wide a spectrum of users as possible it is not unreasonable to dedicate a personal computer to collate the users intentions and link it via an RS232 port using standard ascii characters to a robot controller thereby making for a measure of compatibility between aids. This approach readily allows for a wide range of alternate forms of input that can be combined as appropriate and can be adapted to changes required by an increased familiarity with the hardware or deterioration in the users medical condition. Several current key board emulators already use the RS232 serial interface so could readily become an alternative interface by transmitting the first letters of a command string. Additional information could be gleaned using head or eye pointing techniques or in combination with a vision system to locate approximate spatial positions.

To keep the cost to a minimum and make provision for maintenance it is logical to base robotic aids on an existing robot modified either by the manufacturers or with a suitable kit. These modifications are likely to involve ensuring an adequate level of safety to the users.

Figure 1 outlines the arrangement we are pursuing at Cambridge. A suitable commercial robot is under modification primarily to investigate safety considerations of the aid and the system will eventually include mathematical models of the robot and its likely workspace but it is the user interface that has until now been our prime concern.

BACKGROUND

Speech recognition is acknowledged as a natural fast and simple form of user input and will undoubtedly become an alternative input form. However its present price and reliability prevent its widespread use and this case study is an effort to assess to what extent current commercially available voice recognizers designed for home computers can be used.

MATERIALS AND METHOD

Initial studies are based on the Scott Instruments "Shadow/vet" isolated word recognizer designed for the Apple II computer. Tests were conducted on two days and on both the user was presented with a screen image of an airplane and a landing strip, given the words that would be move the symbol representing the airplane and asked to try to position it on the landing strip.

Although 9 words were available in the first study each user trained between 4 and 7 words depending on ability. On the first occasion a noise canceling head stalk microphone was used and on the second a tie microphone and amplifier. The words used in training were "forward", "left", "right", "back" "pen-up", "pen-down" and "more". "Pen-up" and "pen-down" gave the Logo style game graphics potential and since all commands were used without arguments the word "more" was included so if preceding a word it increase the magnitude of the next command.

CASE STUDY

First trial.
JS, a 14 y.o. spina bifida male who has a normal speaking voice but limited breath control, managed a high level of success when the system was trained with 4 words, a drop in recognition rate was noticed when 2 further words were trained. He managed to maintain reasonably consistent pronunciation between training and using the system and successfully positioned the airplane on all occasions.

GC is a 17 y.o. cerebral palsy with severe spastic quadriplegia whose speech is dysphonic, dysarthric and shows abnormal prosodic features. He trained 4 words and had a low success rate plagued with frequent word misrecognition and word confusion. A sore throat necessitated removing the microphone and head set so that he could drink and when replaced the recognition level dropped noticeably. Frequent retraining was therefore necessary which resulted in a loss of interest and hence a change in voice. The eventual templates were not effective in recognizing his voice and his use of the game was plagued by misrecognized words. His tendency to sigh was often mistaken as an utterance by the recognizer.

DM, a 14 y.o. muscular dystrophy male with a normal but weak voice, should have achieved a high level of success with the recognizer but was not consistent in his pronunciations and the recognizer has no facilities to remove an incorrectly trained template. This resulted in a long training period with a resulting loss of interest and consequent voice changes. Back ground noise may have again aggravated training and the trial was eventually abandoned.

As a result of these trials several short comings were obvious. The noise canceling microphone was totally unsuited as it collected saliva and needed to be removed to enable drinking and eating.

A facility for removing a single bad or false utterance would be useful during the training phase as a sigh was frequently recorded as a word.

Finally the change in interest level between training and use, and during retraining within the game causes a change in voice tone and it is desirable to reduce or eliminate this phase.

Second trial.
A screen identical to that of the first trial was presented to the user. Provision within machine code routines enabled allowed a supervisor to toggle into a training mode where the microphone was disabled and the correct response to the last utterance was indicated before returning to the game.

GC had considerably more success using the recognizer on this occasion and managed to play two games to their completion, something he had not achieved with the earlier versions. The system still misrecognized words but with an almost acceptable frequency. He used the system after a break and a drop in recognition rate was noted.

JS used the game with a high level of success and perhaps was the only student to realize consistent use of words produced better results.

DS, a 16 y.o. muscular dystrophy male with a normal voice that tends to fluctuate in pitch and quality during stressful situations, was initially cautious of the computer having had little contact with computers. His reluctance to use the machine may be attributed to a fear of failure prevalent in muscular dystrophy cases. However he did eventually use the game and achieved a high level of success.

Finally JC, a 19 y.o. quadriplegic cerebral palsy whose voice is dysarthric, dysphonic and has markedly distorted prosodic features, used the game and although recognition was low this was accounted by her extended sounds and her inability to produce the stop to words such as forwar(d) and lef(t). The facility to retrain on the last utterance was, in this case, essential since she is unlikely to have had any success had she used the normal training procedure. In addition a correct response gave her encouragement to succeed and her change in tone was readily accommodated. She would not normally be considered for a robotic aid since she steers her wheel chair erratically but is described here as an indication of the possibilities of this technique.

CONCLUSION

Although a further dimension and hence complexity will be added in a robotic aid two points were indicated by these trials. Commands were given on a trial and error basis and it was noted that more errors were made if the airplane was inverted, in a robotic aid provision must be made to ensure this does not cause a disastrous result.

The facility to train or retrain utterances quickly and inconspicuously during the use of the applications program is likely to be of positive use to the disabled especially those who do not appreciate the details of the technology. Although the superviser intervened frequently in these trials it is hoped that this will reduce once the user becomes familiar with the equipment, with perhaps an occasional need for retraining.

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AN INEXPENSIVE METHOD FOR CONSTRUCTING KEYGUARDS

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ABSTRACT

This paper describes a method for constructing keyguards using clear plastic (plexiglass). Fabrication can be done in a wood shop by individuals who have knowledge of power tools.

INTRODUCTION

A keyguard is a device that allows physically disabled individuals to more accurately access keys on equipment such as calculators, typewriters or computers. Many individuals use the keyguard as a base to rest their wrists on as they manipulate the keyboard with their fingers.

Keyguards are available commercially for popular computers and typewriters for eighty to ninety dollars (U.S.). For individuals who do not have one of the specified keyboard layouts, little is available.

METHODS AND MATERIALS

This technique is useful for flat or nearly flat keyboards.

I

Place the keyboard on a copying machine so that the keys are as close to the glass as possible and make as clear a copy as possible (machines that don't have moving carriages are preferred). Measure the original keyboard length and width then measure the copy. The copy should not be more than 1/8 inch different in size than the original.

II

Cut a piece of plexiglass so that it is two inches longer and wider than the copy. See figure 1. (Plexiglass or lexan is available at most hardware stores.)

III

Plexiglass comes with a protective covering. Do not remove the covering. Glue the copy of the keyboard to the center of the cut plexiglass.

IV

Decide where you will mount the keyguard to the piece of equipment. Many calculators and computers have thin margins of plastic bordering the keyboard layout. Mounting areas should be as long as possible; around the outside of the keyboard area is typical. See figure 1.
Mounting is usually done with Velcro to allow the keyguard to be taken off when necessary. Measure the area the plexiglass will have to cover other than just the keys (mounting area) and mark it on the plexiglass. (Measure from the edge of the keys out to assure proper position.) Cut along the marked lines.

V

With a hot soldering iron mark the center of each key. This acts to center each hole before drilling. Using a 1/4 inch drill bit, rotating at least 1800 RPM, drill each key center. Repeat using a 9/16 inch drill bit or a bit that is of appropriate size for the instrument that will be used to depress the keys.

VI

Round and smooth the top, bottom and inside of each hole using sand paper or a cone shaped grinding wheel. Grinding wheels are available at most hardware stores. Make sure that it will fit your particular drill chuck.

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VII

In most cases the keyguard will be mounted to the surface of the chassis. Since most keys rise up from the chassis you will need to attach a spacer to the keyguard. The spacer also acts to strengthen the keyguard and make it less flexible.

Measure the height of the keys above the surface of the chassis. Add to that 3/32 of an inch to allow for the flex in the plexiglass.

The thickness of velcro (hooks and loops) is between 1/8 inch and 1/16 inch depending how tightly they are pressed together (1/8 inch is a good reference). Subtract this amount from your current dimension. This new dimension will be the thickness of the spacer you will need.

VIII

Spacers can be made out of many different types of materials. Thin strips of plywood or plexiglass are common. Cut the spacer material to fit your mounting area. Place the spacers on the keyboard mounting area then hold the keyguard in place to see that all parts are the correct size.

IX

Cut off all excess plexiglass then remove the protective covering. Attach the spacers to the keyboard. Any type of glue that will bond to plastic is good.

X

Cut velcro to fit spacer and mounting area. Attach velcro hooks to spacer, then loops to mounting area.

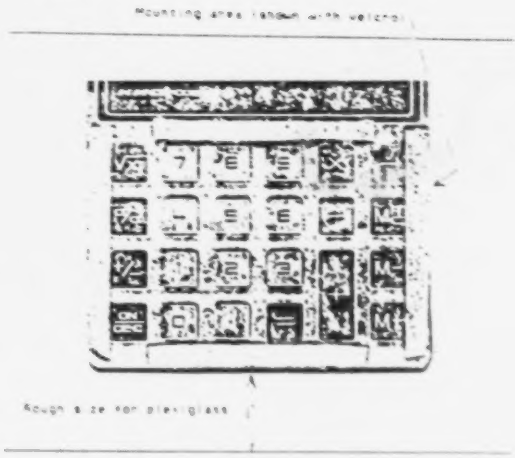
XI

Finally, make sure there are no sharp edges on any part of the keyguard.

CONCLUSION

This technique can easily be adapted for keyboards with slight curves on the key layout or irregular mounting surfaces. It has sucessfully been used to build keyguards for Heath-Zenith, Franklin, Texas Instruments, Radio Shack and Apple computers; Brother, Sears and IBM typewriters and several calculators.

Figure :



ALTERNATE ACCESS TO ALL STANDARD COMPUTERS FOR DISABLED AND NON-DISABLED USERS

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Abstract

Many advances are being made to allow disabled individuals to access computers. Specialized interfaces that can be operated with voice, sip-and-puff, and even the eyes have been developed. For blind individuals, special output forms using braille and voice have been developed. Unfortunately, most of these adaptations could originally be used only with specialized or selected software packages. An "alternate access" approach is now being developed which would allow disabled individuals to access all standard software programs as well as the specialized software. In addition, these "alternate access" techniques promise to benefit non-disabled users as well.

The Triple Roles

In examining these problems, it is important to realize that there are three roles that computers play in the rehabilitation of disabled individuals. The first role is as a therapy device. Therapy for many disabilities, especially those from brain injuries, may be able to take advantage of the low cost and extreme patience of computers to provide individuals with self-paced therapy programs.

A second application of computers is as personal aids to disabled individuals. Here, the computer running special software could act as an individual's "pencil and paper," filing system, translator (translating text into speech or braille), etc. In this role, the computer acts as a personal aid to the individual, and provides functions which other individuals generally achieve in some other fashion. Such a computer is owned by the individual, and can have special programs and modifications made for it.

The third role that computers will play in the lives of disabled individuals is the same role that they will play in the lives of non-disabled individuals. Computers are very quickly being incorporated into our society at all levels, from education to employment to daily living to recreation. These computers generally do not belong to us, but are rather computers which we encounter in the environments of our schools, jobs, etc. It is this third role of the computer that we are concerned about in this paper. Specifically, we are concerned about the severe handicap that a disabled individual will experience in education, employment, and even daily living, if he is not able to freely access and use the many different computers which he is going to be encountering in his daily life. In fact, if access to these different computers cannot be assured, participation by the disabled individual in most regular education and white collar employment settings will become impossible within the near future.

The Solution Strategy

The solution strategy proposed and being explored is to put a generic input/output port on all computers and information systems. This input/output port would accept input from alternate input devices (special keyboards, special aids, etc.) and treat it exactly as if it came from the standard input devices (the keyboard, a mouse, a touchscreen). A similar treatment would also be made for any information being sent out to the standard output display. Here, any information being sent to standard output displays (CRT screens, speakers, printers) would also be present on the alternate output connector, which could be connected to alternate displays (braille displays, voice output, etc.).

Such an approach would not only allow use of the system by disabled individuals, but would also enhance use by non-disabled individuals.

A disabled individual would use the system by simply plugging his own specialized input system(s) into the jack, and proceeding to use the system as would anyone else. A disabled individual would have to secure for himself specialized input/control systems optimized for his residual abilities. These communication aids or interfaces would be kept by the individual as a personal aid. Whenever he encountered an information system he needed to operate, he would simply connect his interface and use it.

Non-disabled users could use the alternate inputs to connect their own specialized inputs for various purposes. Individuals who like to use specialized one-hand keyboards, Dvorak keyboards, abbreviation expansion, or shorthand keyboards, etc., could simply connect them to whichever computers they were using given time. Manufacturers would also be able to make generic "voice input keyboards" which would connect into any company's computer, since the "alternate access" input port would be standardized. The alternate input connector would also facilitate use of the computers in environments where specialty keyboards were required. For example, if a computer was needed in an explosive or hostile environment, a "standard hostile environment keyboard" could be connected to the computer on this alternate access port.

The alternate output portion of the connector would function in a similar manner. Blind or other sensory-impaired individuals would keep special tactile, vocal, or visual (for the deaf) displays with them, which could be connected to the computers and information systems they encountered in their environments. Any information which was produced in a form that they would not ordinarily be able to perceive could be translated by their personal device into a form usable by them. Again, the non-disabled market could also make use of these output connections to provide generic input to special displays of many types.

Work to Date in this Area

Work on this overall objective is currently proceeding in both the disabled and non-disabled market. In the disabled areas, a small device called the keyboard emulating interface (Vanderheiden, 1981; Korba et al., 1983) has been proposed and developed for a number of popular computers. This module is installed in the computer between the standard keyboard and the rest of the computer. It accepts keystrokes from either the standard keyboard or a special RS232 input and passes them on to the computer as keystrokes. As a result, information fed in through the RS232 port of the emulator is passed on to the computer in precisely the same form as keystrokes from the computer's standard keyboard. The computer cannot tell keystrokes coming from the keyboard from those coming in the serial port. The keyboard emulating interface therefore allows individuals using specialty input devices to use the computer and all of its standard software. Any software which operates off of the keyboard can be used by the disabled individual without modification. This type of modification is termed a "transparent" modification.

Transparent techniques are also being developed for blind and visually impaired individuals. For these individuals, the problem is dealing with the CRT screen. Techniques have been developed which tap into the operating system of the computer and are able to track the information which is written to the screen. This information is then displayed on a special display (voice, braille, or expanded display) usable by the visually disabled person. The system from Visualtek is a good example of a system for individuals with low vision (Vanderheiden et al., 1984). Their aid (which has versions for the Apple, IBM, Radio Shack and other computers) recreates the characters in an enlarged format on a separate display. The size of the characters can be varied, and can be made so large that a single character fills the entire screen. Using a joystick or other interface, the individual can have the text automatically scrolled past at a user-controlled rate. Since the system is completely transparent, and does not interfere with the operation of the standard computer, it can be used with standard off-the-shelf software.

For individuals who do not have residual vision, other systems are available which provide transparent voice output of the content of the computer's screen. Systems such as Blind.talk, for the Echo II or Echo PC, and Total.talk, for the IBM, allow the disabled user to use the standard unmodified software and have the computer read out the contents of the screen at the user's direction.

All of these systems have their limitations, however. None of the computers or operating systems were originally designed to support these types of modifications. As a result, the modifications must be custom designed for each new computer, and each new model of a computer, as they come out. This is costly, and does not allow the user to easily move between brands of computers, taking their special hardware with them. Also, as new operating systems are introduced or updated, additional time must be spent on the part of the manufacturers of the special interfaces to ensure that their modifications will still work, or to revise them so that they will. Provision of a standardized access port to the keyboard handling and display driving routines would greatly simplify the task of these specialty manufacturers, and increase the longevity and applicability of their devices across computers, benefiting both manufacturers and consumers.

Opportunities from Advances and Directions in the Non-Disabled Market

Although such a standardized access system would be beneficial to disabled individuals, it normally would be unrealistic to expect that such a standard access port be adopted or implemented on standard manufacturers' computers. There are, however, a number of developments in the field of microcomputers which are heading precisely in this direction.

A number of operating systems are picking up the concept of a "keyboard object." For example, TopView, the new operating environment from IBM, is capable of running several programs concurrently. Since they are not all connected to the keyboard at the same time, TopView sets up a "keyboard object" for each program. Any characters put into the keyboard object are used by the program as if they came from the standard keyboard. One result of this is that one program can in fact create the "keystrokes" for another program. All of the programs running in the background of TopView receive their input through such a keyboard object. If this can be extended to the foreground program as well, and an option built into the system to allow it come from an external input port, then the problem of handicapped access to the standard keyboard would essentially be solved (for this case). Since a good case can be made for this for the non-disabled market (to allow use of alternate keyboards, voice input, etc.) as regular computer inputs, the concept becomes feasible. In fact, similar capabilities have been available on mainframe computers for some time.

Similarly, work with virtual display interfaces (VDIs) may provide the capabilities and links to the standard visual display necessary for alternate displays. The purpose of the virtual display interfaces (which again is being developed for the standard computer market) is to allow a program to create a visual image on a screen with unknown resolution; that is, the program can write characters, draw circles, etc., on a screen when it doesn't know exactly how many dots there are, or exactly what the screen configuration is, in advance. Instead of sending the dot information directly to the screen, therefore, the program passes the information to the VDI, which connects to routines that draw/write the information on the screen with the maximum resolution available to that screen. Since an individual may want more than one display connected to a computer (e.g., a built-in screen and a high-resolution projection screen), there may also be need for exporting this VDI information so that it can be sent to the separate projection display. Again, the availability of the video display information in standardized form, on an external connection, could provide the groundwork for the use of alternate displays for disabled individuals.

Pipelining, new windowing capabilities, and the increasing use of multitasking and nesting in the newer computer systems is therefore beginning to provide us with exactly the tools we need (Vanderheiden, 1981) for making computers more accessible to a wider population. Moreover, these developments are occurring within the standard computer field, and for the benefit of the non-disabled population. If these developments are implemented properly, they will go a long way toward assuring alternate access to standard computer systems for individuals with disabilities as well as access for the general population. If, however, they are not implemented properly, we may have the equivalent of a situation where an entire city is paved with nice concrete paths which greatly facilitate the mobility of disabled individuals, but which all have a 6" or 8" curb at the corners.

In order to help explore the needs of disabled individuals, and to be sure that unnecessary (and unneeded) curbs are not put in accidentally, a number of efforts have been started to bring the manufacturers of standard computer hardware and operating systems together with researchers, consumers, and developers of specialized access hardware, to see how the new developments such as those discussed above can be implemented in such a way that they maximize the potential and opportunities for individuals with disabilities. Several projects at the Trace Center are now focused on working with standard hardware manufacturers to facilitate this process. The Office of Special Education and Rehabilitation Services (OSERS) of the Department of Education has also instituted a major initiative in this area, which is being coordinated with the White House Private Sector Initiatives Program. Through these and other cooperative efforts nationwide, there is reason to be optimistic that the type of solution strategy being proposed can and will be realized. The most compelling reason for this optimism is that it is likely that more non-disabled individuals will benefit from it than disabled individuals. As a case in point, curbcuts are used by many, many more non-disabled individuals than disabled individuals.

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18.1

An Experimental Study of Knee Kinematics In-Vitro.

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ABSTRACT

A rig to analyze 3-dimensional knee joint kinematics In-Vitro was designed and constructed. Criteria for design included 1) full six degrees of freedom, 2) data acquisition during dynamic flexion-extension under quadriceps action, 3) output in Eulerian angles and translations referenced to reproducable axis system, and 4) ability to measure neutral path motion and "kinematic envelope".

Seventeen specimen were tested. Motion paths were consistent between specimen, with clear "screw home" mechanism. The ACL was sectioned on twelve of the specimen and re-tested. Motion pathways were again consistent, however a large pivot shift (posterior subluxation with flexion) was evident.

INTRODUCTION/BACKGROUND

In the past, studies concerning knee motion have been limited in a number of ways.

The simplest studies rely primarily on measurements made by hand with a clinical goniometer (1). Complex analyses have also been made from mathematical models which are purely theoretical (2,3). More recently, gait analysis has been applied to the study of knee motion (4). There is, however, questionable correlation of leg kinematics and bone kinematics in these studies.

Perhaps the most popular technique, which has been employed with reasonable success in a number of motion studies (5,6,7) is a radiographic method which uses a rig for knee loading and positioning. These studies are usually on a static knee set to discrete flexion angles. Further, these planar analyses rarely account for the 3-D mechanics of the knee joint.

Electrogoniometers have seen frequent use in the study of knee kinematics (8). While this provides for dynamic measurement, attachment to the specimen is a problem. Also, few, if any goniometric studies have allowed measurement of all six degrees of freedom.

Another common problem in kinematic research is the lack of a consistent reproducible axis system. Without this, the correlation and comparison of results from one investigator to the next is difficult, if not impossible.

This paper describes a dynamic test rig which overcomes these problems. It allows measurement of both neutral path motion, and laxity through internal-external biases and shear across the joint. Experimental results are presented for seventeen knees tested using this rig.

METHODS

Coordinate System

The fixed axis is in the tibia and the moving axis in the femur with the two systems coincident at full extension. Knee geometry is the basis for axis location (9): The posterior femoral condyles are nearly spherical surfaces and the femoral x-axis passes transversely through the center points of these spheres (positive is medial for a right knee). The y-axis is parallel to the long axis of the bone (positive upward), and the z-axis is mutually orthogonal (positive anterior).

A 300 mm long rod is placed through the femoral condyles along the transverse axis using radius gages and a drilling jig. This rod defines the axis system and is used to accurately position a bracket onto the femur, then to position the tibia within the base of the test-rig where it is fixed with acrylic cement.

Drive System

The knee is cycled through a flexion-extension range of approximately 120 degrees during testing. Attached to the femoral bracket is a 4 kg lead weight located 40 cm proximal and 10 cm posterior to the transverse axis. This weight produces a flexion moment on the knee. A sheathed cable (bicycle brake cable) is attached to the femoral bracket with one end clamped to the quadriceps tendon through a load cell, and the other end wound around a spool. The spool is driven by a programmable stepper motor which flexes the knee dynamically between two set points. The rate of flexion can also be varied.

Measurement System

An electro-goniometer is mounted on the femur parallel to the transverse axis to monitor the flexion angle. Three Celesco PT101 cable-type position-displacement transducers attach to each end of the 300 mm long transverse axis to measure the 3-D orientation. At full extension each set of transducers is parallel with the three coordinate axes with known initial displacements from six reference points (figure 1). In the flexed configuration the x, y & z-coordinates of the endpoints of the axis can be calculated knowing the new transducer displacements (10).

Data Acquisition & Manipulation

The knee test rig is hooked into an Apple IIe computer via Mountain A/D card. Although the knee is flexed and extended in a continuous cycle, data is collected only at discrete 5 degree increments. The six transducer displacements for each flexion angle are transfered to a VAX 11/750 computer where Euler rotations and translations are numerically computed. Graphs of these results are displayed on a Ramtek 6221 Color Graphics terminal. Display can be produced for a) a single run, b) the range over multiple runs or, c) means & standard deviations over multiple runs.

Variations

In addition to neutral path motion, three biased configurations were used a) posterior shear of the femur on the tibia obtained by angling the entire apparatus by 20 with respect to the vertical, b) internal rotation of the femur on the tibia with constant torque of 1 N-m, c) external rotation of the femur on the tibia with constant torque of 1 N-m.

Specimen

Seventeen fresh frozen knees were tested twelve of which were tested a second time after sectioning the anterior cruciate ligament. Specimen were thawed overnight and dissected down to the joint capsule leaving intact the entire capsule, quadriceps tendon, and fibula. All experimental runs for a given knee were performed in a single day. The ACL was sectioned by making a small incision into the joint through the fibers of the patellar ligament-- otherwise the capsule was not disturbed. Following testing the joint was opened to verify that the ligament was successfully sectioned (hence only 12 of 17 specimen had valid ACL deficient data).

RESULTS

A graph of normal motion averaged for the seventeen specimen is shown in Figure 2. The predominant aspects of knee kinematics are: 1) 11.5 degree internal rotation of the tibia with flexion, 65% of which occurs in the first 30 of flexion. 2) Y-displacement of the femoral origin by -6.3 mm (laterally) with flexion, 69% in the first 30 of flexion. 3) Z-displacement of the femoral origin by -4.5 mm (posteriorly) with flexion, 96% in the first 30 of flexion.

The kinematics are relatively insensitive to variations in load across the joint, posterior shear on the joint, and speed of flexion-extension cycle. The addition of biasing torques cause the internal rotation curve to shift by a constant amount: -2 for a 1 N-m. external torque, and +2 for a 1 N-m. internal torque.

Typical loads on the quadriceps tendon were 200 N at full extension increasing to 600 N at 80 flexion and then decreasing to 450 N at full extension. Quadriceps forces averaged 25% higher on the extension cycle compared to the flexion cycle.

For the ACL deficient knee, plots of internal & varus rotations, and x & y-displacements were similar in shape and magnitude to those of the intact knee. Plots of z-displacement, however, were markedly different from those of the intact knee (figure 3). Z-displacement of the femoral origin was -9.7 mm (posteriorly) in the first 30 of flexion and then moved anteriorly to -5 mm at full flexion. Additionally, this z-displacement was sensitive to posterior shear on the joint with the maximum z-displacement shifting -4 mm upon the addition of shear forces.

The "kinematic envelope" for a typical knee is shown in figure 4 for the intact knee and figure 5 for the ACL deficient knee. These plots include both neutral path and biased motion. The maximum laxity for this knee is +3.5 for internal rotation, +0.4 for varus rotation, and +1.4, 0.2, 1.6 mm for x,y,z-displacements respectively.

Conclusions

All experimental runs demonstrated the "screw home" phenomenon i.e. internal rotation in the initial phase of flexion. There was little variability in the average motion of the knees. However, ACL deficient knees experienced a large posterior subluxation of the femoral origin in the intitial phase of flexion which was not present when the ligament was intact. Motion envelopes were fairly small suggesting that motion outside these natural boundries may be deleterious to the soft tissues of the knee (ligaments and menisci).

Future applications of this rig would be the design and testing of total joint replacements and the design of knee braces. In addition the rig can be used to analyze the role of the various soft tissue structures of the knee on kinematics. This has particular importance to ligament reconstruction, and the design of surgical procee-dures.

ACKNOWLEDGEMENTS

We wish to thank Jorie Ewald, Peter Chow, Phil Carbone and Anna Sawczuk for their contributions to this work. We thank Harvard Anatomy Department for providing specimen.

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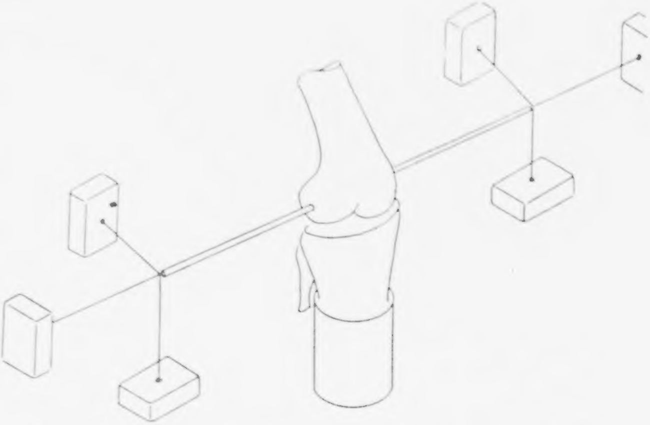

FIGURE 1. Schematic representation of knee test rig. Shown is the transverse axis through the femur attaching to two sets of position/ displacement transducers. The transducers and the tibia are fixed to the base.

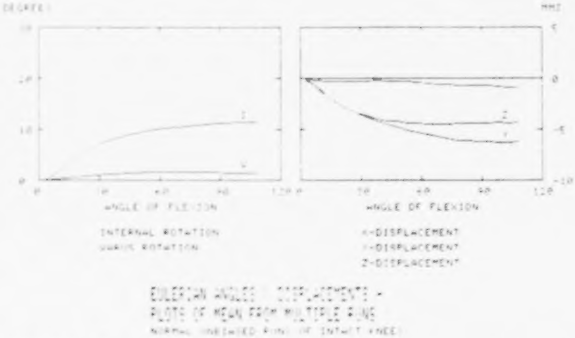

FIGURE 2. Neutral path of motion averaged for seventeen knees with intact ACL.

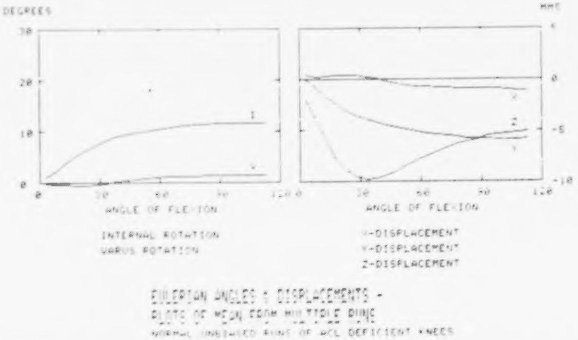

FIGURE 3. Neutral path of motion averaged for twelve knees with sectioned ACL.

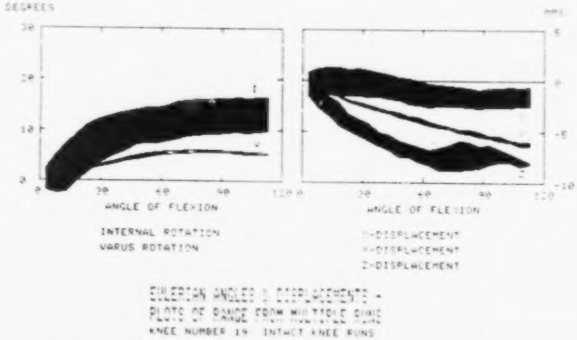

FIGURE 4. Typical "kinematic envelope". Shown is the entire range of motion resulting from loading biases on a knee with intact ACL.

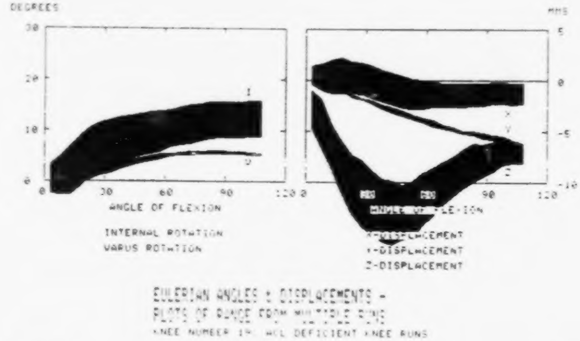

FIGURE 5. Typical "kinematic envelope". Shown is the entire range of motion resulting from loading biases on a knee with sectioned ACL.

The Effects of External Knee Joint Motion on Internal Knee Mechanics

18.2

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INTRODUCTION. Orthotic devices which use fixed axis or polycentric external joints, are widely used in the treatment of joint motion restoration, in post-operation and post-injury ligament protection and in ligament injury prophlaxis. In these applications the external device guides the motion of the knee. Abnormalities of motion will be conteracted by soft tissue deformation, cuff-skin slippage and ligamentous tensions. In addition, meniscal location, and other aspects of joint mechanics will be abnormal. Therefore it is important that the imposed motion pattern is as accurate as possible. It is assumed that the most suitable motion for an off-the-shelf orthotic is 'average knee motion' (2). This study examines the effect of an external joint providing 'average knee motion' on the ligament length patterns and meniscal location of individual cadaver knees. The effects of deviations from average motion are then examined.

METHODS & MATERIALS. Twenty-three fixed cadaveric knees with no significant abnormality were obtained from the Harvard Anatomy Department. Based on the finding that the posterior femoral condyles could be closely represented by spherical surfaces (1,2), a transverse axis through the medial and lateral sphere centers was located with a Steinman pin, using special radius gages. The femoral and tibial insertion points of the anterior and posterior bands of the ACL, PCL, and MCL ligaments were marked with metal pins. A single band was marked for the LCL. The knee was positioned at 0 degrees flexion and placed in a jig with the transverse axis located at a right angle. Biplanar x-rays were taken, the ligament insertion points were digitized and then the actual coordinates were computed using trigonometric relations. The knee was then embedded in Pedelin foam, and then 24 equal sagittal slices were cut on a band saw. The 25 resulting sections were reproduced at 1:1 and the outlines of the femur, tibia, fibula, patella and meniscus, were digitized into the computer. All coordinates were then standardised to an average width across the transverse axis of 80mm (3). The knee was then 'flexed' in the computer according to average knee motion. The flexion angle was specified, and then varus-valgus, int-ext rotation, a-p displacement, m-l displacement, and dist-prox displacement are imposed by a transformation matrix as Eulerian angles and displacements, as function of the flexion angle (2). The computer program then determined if the lateral and medial condyles were in contact within 0.5mm and if not, corrected in the frontal tibial plane by a varus-valgus rotation and a dist-prox displacement. The lateral and medial gap discrepancies, and the coordinates of the femoral-tibial contact points, were recorded for all flexion angles from 0 to 120 in steps of 15 degrees. At each angle of flexion, when the gaps had been corrected, the transformation matrix was applied to the femoral ligament attachment coordinates, and the new ligament band lengths were computed. By altering the equations for knee motion, all of the data was obtained for int-ext rotation eliminated (resembling a so-called polycentric joint), for a-p displacement eliminated, and for rotation and a-p eliminated (close to a fixed axis joint). All the various positions of the femur were checked with color graphics displays, while ligament length patterns were also displayed and checked graphically.

Results The mean and standard deviations of the results were computed for the group of 23 knees. The lateral and medial gap discrepancies, due to imposing 'average knee motion' on the group of 23 specific knees, are shown in figures 1 and 2. The largest discrepancies, averaging 1.3mms, were on the lateral side, particulary in flexion (fig. 1). On the medial side, the gaps averaged only about 0.5mm throughout motion. This indicated the inherent validity of the average knee motion equations. For all of the motion, the length pattern of the collaterals remained relatively unchanged: the LCL slackened by about 20% during flexion; the MCL remained within 15% of its zero length. Large changes were seen in the cruciates however. Compared with 0 flexion, the anterior band of the ACL loosened about 20% and the posterior band loosened then returned to its 0 flexion length during flexion. The posterior band of the PCL minimally loosened throughout flexion while the anterior band tightened a striking 40% (fig. 3). The data is comparable to that of other reports (4). Large changes in these patterns produced by the different motions was noted (fig. 4). However, the largest changes occurred when a-p motion was eliminated. The center of the meniscal location was defined at the femoral-tibial contact points. In 'average knee motion', the lateral center moved steadily posteriorly by about 20mm. The medial center moved back about 10mm from 0 to 30 flexion and then remained relatively static (fig. 5). (The medial meniscus probably remains static even from 0 to 30, because of the femoral condyle rolling on the anterior surface of the meniscus). There were serious abnormalities in the meniscal locations for all of the abnormal patterns of knee motion (fig. 5).

CONCLUSION 1. Imposing 'average knee motion' on 23 knee specimens resulted in only small errors in femoral-tibial condyle contact.

2. Mean ligament length patterns for average knee motion were determined. Large changes resulted from imposing abnormal knee motion, especially the elimination of a-p displacement. 3. Abnormal meniscal locations were imposed for all of the abnormal knee motions.

APPLICATION The data has been used to design an external joint, to be incorporated in a brace (5). The joint is designed by computergraphics and made by an NC machine, to produce 'average knee motion' or any desired motion for protection of specific ligaments. The braces are undergoing clinical evaluation.

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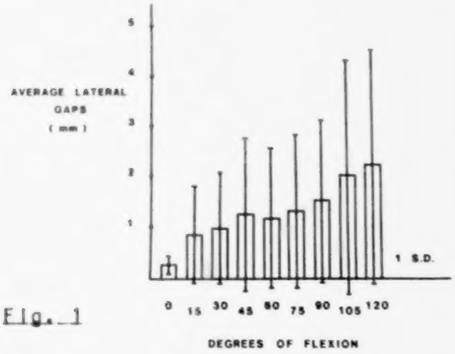

Fig. 1

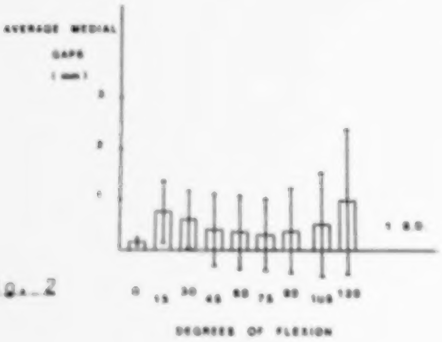

Fig. 2

LIGAMENT LENGTH PATTERNS

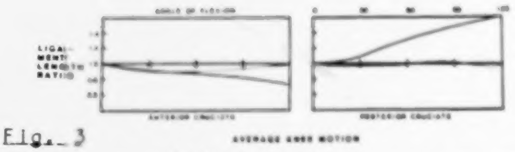

Fig. 3

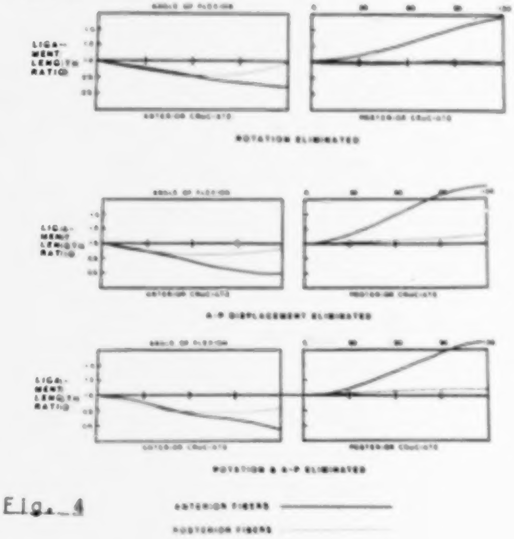

Fig. 4

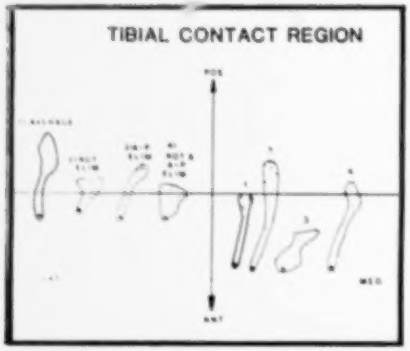

Fig. 5 A-P displ. of fem-tib contact points for 4 motions from 0 to 120 of flexion. Large ant. dots denote contact points at 0 flexion.

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PRESS-FIT TOTAL KNEE REPLACEMENT

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ABSTRACT

Criteria for the design of the bearing surfaces and the interfaces are discussed. The surfaces should provide a "kinematic envelope", to allow a variety of motion paths and minimize interface stresses. The interface should transmit stresses to the bone surfaces as physiologically as possible. Contact pressure studies and finite element analysis showed the stress distributions on the tibial surface with different interface conditions. A Press-Fit Total Knee was designed and clinical evaluation has been in progress since February 1984. This paper will present results of the studies and discuss the viability of the concept.

INTRODUCTION & BACKGROUND

Most total knees in use today are designed to be fixed to the bone using acrylic cement, or by bone ingrowth into porous surfaces. Cement has excellent results in the medium term but may have long-term problems. Tissue ingrowth, while promising, is relatively new and unproven. A third alternative is to interface the metallic prosthetic components directly against the bone, called "Press-Fit". the potential viability of the concept has been proven in long-term results of devices such as the MGH femoral component, McKeever tibial plateaus, and Ring femoral endoprostheses. Advantages include simplicity, low cost, and ease of removal.

It is proposed that the stability of the interface depends upon the stress distribution on the resected bone surface -- the more physiological the stress distribution, the greater the durability and vice versa. The factors affecting the stress distributions are:

1. Femoral-tibial bearing geometry. Low conformity reduces the shear forces carried by the component, the main trade-off being higher metal-plastic contact stresses and possibly higher wear (1, 2).
2. Location of femoral-tibial contact points. This is affected by component geometry, retention or sacrifice of cruciate ligaments, and surgical technique (3).
3. Overall leg alignment, although the direction of the force vector in the frontal plane cannot be accurately predicted from static alignment (4).
4. Component coverage of the bone surfaces, and real area of implant-bone contact (5).
5. The material comprising the interface at surgery and after biological remodelling (6, 7).

Given that all of these factors cannot consistently be optimized, our goal was to design an interface with the maximum margin for error. The possibilities of a metal-bone Press-Fit were analyzed, and the enhancing effect of a compliant layer between metal and bone was considered.

MATERIALS AND METHODS

a. Finite Element Analysis

A 2-D plane strain ANSYS analysis model was defined at the centers of the lateral and medial plateaus in the saggital plane, of a normal and an arthritic specimen. The geometry and the elastic modulus were determined by CT scans. An experimental correlation between CT number and maximum elastic modulus E taken along the principal trabecular orientation (8) resulted in two best fit quadratics:

-100 to 400: $E = 55.45 + .149\ CT + .00093CT2$
400 to 1000: $E = 3995 - 19.55\ CT + .0255CT2$

A rigid boundary constraint was defined for the lower 25% of the section (Fig. 1). A force of 100 N/mm was applied to all models. Physiological loading was simulated by a parabolic load over the posterior 2/3 of the section (9). The tibial component was 5 mm of polyethylene (1E3 MPa) on 3 mm of metal (2E5 MPa). The load was applied on the plastic along a 10 mm width in the center (3). the intervace between the metal and the bone was 1 mm cement (2E3MPa, $v=.23$), 1 mm compliant material (1.9MPa, $v=.495$), or direct metal-bone contact. Each model had 1200-1800 linear 4 node Stiff 42 one mm square elements. Triangular elements were used at the edges. Linear static analysis was done on all models.

b. Determination of Real Contact Area

The upper surface of 11 tibias was resected, and the tibias were mounted vertically in a load machine. Loads up to 3000 N were applied to the resected surface through a flat rigid block. Fujifilm pressure sensitive film against the bone surface recorded the loading points, which were analyzed in an optical imaging system. Three interfaces were tested on each bone, rigid, rigid + compliant, rigid + cement. A further 10 tibias were tested, the interfaces being rigid, 1 layer compliant, 2 layers compliant. Finally cyclic loading tests at 2000 N were carried out to study trabecular fatigue and changing contact patterns. Five were tested with a rigid interface, and three pairs compared rigid with compliant.

RESULTS

For the intact tibias, the highest stresses in the bone were posterior (Fig. 1). the stresses beneath the implant were more uniformly distributed, but had higher anterior and lower posterior stresses compared with normal (Fig. 2). Resection of 5 mm of bone gave equal or even lower stresses compared with the minimal resection condition. This was due to increased surface area after the resection, especially laterally. Rigid and cemented interfaces gave virtually identical stresses under all conditions. The compliant layer "smoothed out" the stresses, and reduced peak stresses by up to 50 percent in regions of high elastic modulus gradient (Fig. 3).

However, incomplete surface contact was found for rigid and cemented interfaces in the tests. Contact was expressed as the number of pixels which showed contact on the pressure patterns (Fig. 4).

Interface	No. of Bones	Mean Contact Pixels	Std. Dev.
Rigid	11	1645	543
Cement	11	1399	419
Compliant	11	2500	1038
Rigid	10	2298	644
1 Compliant	10	3170	804
2 Compliant	10	4140	1124

In the 5 tibias cyclically loaded, the component subsided more than 2 mm in 2 bones, subsided < 1 mm in 2 bones, with no change in 1 bone. The pressure patterns showed that load had transferred from the interior to the periphery after cyclic loading, but had not become more uniform or extensive. For the cyclic rigid vs. compliant, one pair showed massive subsidence of the rigid side at 50,000 cycles, while the compliant side showed less subsidence at 150,000 cycles. Another pair had minor subsidence and changes in the pressure pattern with the rigid interface, but no changes with the compliant side. The third pair did not demonstrate subsidence or change in pressure patterns.

On the basis of the above, a further finite element study was carried out with incomplete surface contact. This demonstrated abnormal stress patterns in the surface layer which could lead to bone resorption in regions of zero and high stresses (Fig. 5).

Based on the clinical history of Press-Fit components alone (cited in intro), and based on optimization of as many of factors 1-5 (above) as possible, a Press-Fit Kinematic II knee was designed and manufactured. At this time (Jan 85) it has been used in 5 patients for specific indications (Fig. 6). The surgical recovery has been similar to that for patients with cemented total knees. the patients are being followed functionally and radiographically. Animal studies (6) are being analyzed to test the hypothesis of further improvement of the interface with a compliant velour.

CONCLUSIONS

The location of femoral-tibial loading, the level of tibial resection, and the nature of the implant-bone interface, were shown to affect stresses in the surface layer of bone. A compliant interface layer significant reduced peak stresses in regions of high bone modulus gradient. Actual implant-bone contact areas were found experimentally to be much less than apparent contact areas. Analysis showed this resulted in regions of high and low bone stresses. This research is being used in the understanding and design of Press-Fit Total Knees.

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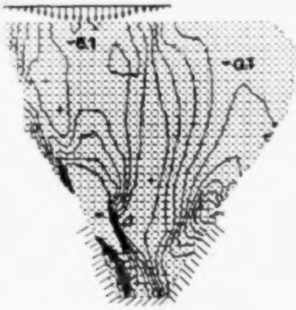

FIG.1 Boundary constraints (lower 25%), loading profile (posterior 2/3) and lines of equal compressive stress, for physiological loading of the medial normal tibia.

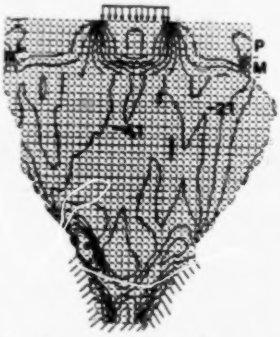

FIG.2 Loading profile (central 10mm) and lines of stress for loading of the tibial component, medial normal tibia.

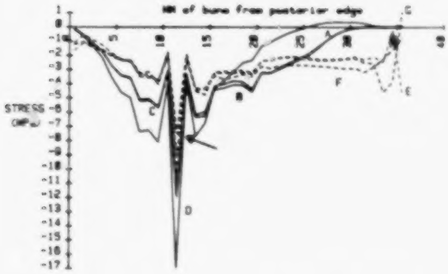

FIG.3 Stresses in the bone just below the component. The compliant layer (arrow) smoothed peak stresses compared with physiological (D) and rigid interfaces (A,C).

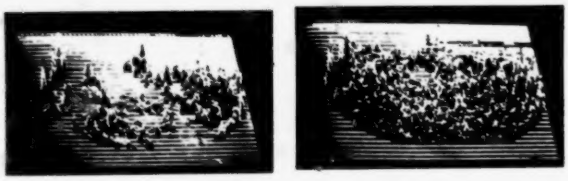

FIG.4 Contact pressure patterns on the resected upper tibia for a rigid interface (left) and with an interposed compliant layer (right).

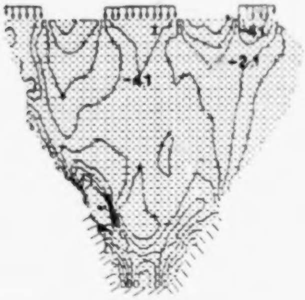

FIG.5 Local areas of abnormally high and low stresses can result from incomplete complete contact of a rigid interface.

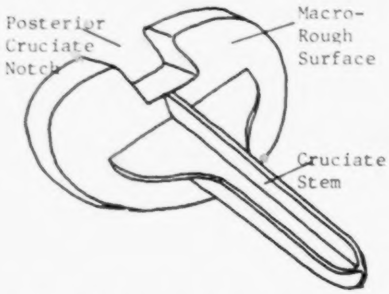

FIG.6 Bottom view of tibial component of press-fit knee.

FIGURES FOR P.S. WALKER, PH.D. ET AL, PRESS-FIT TOTAL KNEE REPLACEMENT.

18.4

JAIPUR AK PROSTHESIS: RESULTS OF MATHEMATICAL MODELLING AND FIELD TRIALS

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ABSTRACT

Conventional Western AK prostheses do not suit most of the amputees in India especially because squatting and floor sitting are very common. A simple endoskeletal AK prosthesis has been developed using local materials and craftsmen. The limb has been tried out on 200 amputees and its theoretical limits of function evaluated with the help of a mathematical model. The results indicate that it is possible to optimize a simple prosthesis which is socially acceptable and provide near normal gait on level ground at walking speeds around 1.1 m/s. For a given mass, geometry and location of center of mass of the shank the walking speed cannot be varied very much. The timing and inclination of the footpiece at toe-off largely determines the swing-phase of the limb.

INTRODUCTION

The conventional above-knee (AK) prosthesis poses two major problems for the Indian amputee. The usual pelvic suspension system, with its metallic hinge joint at the hip, does not permit abduction and external rotation which is essential for cross-legged sitting on the floor. The rigid thigh and leg pieces of the exoskeletal limb strike against each other when attempting full knee flexion and so the amputee cannot squat. In squatting full knee flexion is made possible because the soft tissues of the thigh and knee can be pressed and flattened against each other. Since a Western amputee does not usually assume this posture, the knee joints of most prostheses do not cater for full knee flexion.

The Jaipur AK limb was therefore designed to permit these postures so essential to the Indian life style where floor sitting is a social norm. The rigid metallic hip joint was replaced by flexible leather straps for suspension, allowing freedom of movement in all directions; a posterior elastic strap was strategically located to tighten when sitting cross-legged, thereby keeping the socket pulled up snugly against the stump.

To permit squatting, the endoskeletal concept was utilized. Its soft foam covering can flatten when pressed during full-knee flexion. A new design of the knee joint was worked out. It not only allowed for a full knee flexion but also had an offset hinge to provide alignment stability during the stance phase of walking. An additional locking system was provided as a safety measure. It may be emphasized that the endoskeletal concept was not used as a labor saving device but primarily to allow amputees to squat.

Aluminum sheet was used to fabricate classical quadrilateral sockets and conduit steel pipes for load bearing. The offset single-axis knee-joint and the Jaipur foot (1) are fixed to this pipe using a shrink fit technique.

A saucer-shaped aluminum alignment disc is fitted at the top and a static alignment is secured by placing this disc below the socket and temporarily brazing the two together at a few points. The patient is then made to walk and adjustments in alignment are made if required. Once a satisfactory fit and alignment are obtained, the alignment disc is securely welded to the socket. A foam covering and a special latex-treated cloth cover is then used to provide a proper shape to the limb with a waterproof tear-resistant skin which can withstand the rough exposure of a rural environment.

The theoretical limits of performance of this AK prosthesis were evaluated with the help of a mathematical model for the swing-phase and limbs were fitted to 200 amputees also for field evaluation.

MATHEMATICAL MODELLING

Methodology

A number of studies (2,3,4) have shown that an unconstrained pendulum can approximate the swing-phase motion of the shank since muscles do not play a significant role except in the initial and final stages of the swing-phase (2,4,5). Therefore in this study a mathematical model based on pendulum motion was used to simulate the swing-phase of the prosthesis under different conditions. The objective of the study was to determine the optimal location of the center of gravity of the prosthesis and also its limitations under different conditions of walking. For the purposes of this analysis the use of the Jaipur Foot and mild steel conduit pipes were taken as given. The nominal mass of the shank and foot was 1.4 kg. It was assumed that extra mass could only be added along the shank of the prosthesis and a limit of 1.5 kg was fixed for this extra mass M so that the total mass would not exceed 3.0 kg.

The shank was assumed to behave like a pendulum free to rotate about the knee joint. It was also assumed that the prosthesis knee downwards was a rigid body which moves through space in the swing-phase under the influence of gravity and a specified trajectory of the knee. The trajectory of the knee was obtained from literature (6) for 'Normal' gait on level ground at 95 steps per minute at a speed of 1.1 m/s.

It was assumed that the amputee can move the stump like a normal person hence providing a 'normal' knee trajectory for the prosthesis. Then the equation of motion (Langrangian) for the shank in the swing phase with a frictionless knee-joint can be shown to be

$$\ddot{\theta} = - \frac{Wr}{Io} (\ddot{Y} \cos \theta + (\ddot{Z}+g) \sin \theta)$$

Polynomial curves were fitted to the knee displacement data using a least square technique and integration of the above equation was done numerically using the Merson's form of the 4th order Runga Kutta method. All computations were done on an ICL 2960 computer.

In order to check the sensitivity of shank motion to friction at the knee joint a few runs were made assuming different values of the coefficient of friction. The effect of friction was found to be negligible for values of coefficient ranging from 0 to 0.4. This is so because the diameter of the pin is very small and so the moment generated by the mass of the limb is much larger than the moment due to friction.

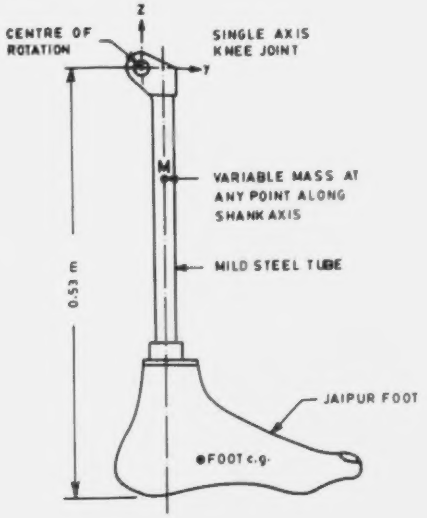

FIG.1 LOWER LIMB OF A SIMPLE ABOVE KNEE PROSTHESIS WITH SINGLE AXIS KNEE JOINT. MASS OF LEG WITHOUT M IS 1.4 kg

Therefore, for all other analyses in this study the knee joint was assumed to be frictionless. No knee-stop was provided in the model and so the shank was free to assume a hyper-extended position. Each run was terminated at the time of expected heel-strike. Shank position was determined by calculating the shank angle with the vertical at different points in time from toe-off to heel strike.

Results

The shank motion was found to be very sensitive to the value of angular velocity at toe-off and instantaneous values of vertical acceleration of the shank. The motion of the shank was less sensitive to the location of the center of the mass. Given the constraints of the geometry of the prosthesis and limitations on maximum allowable mass, it was found that only small variations can be made in the swing-phase motion by altering inertial properties alone.

The results show that for a given mass, geometry and location of center of mass of the shank the walking speed cannot be varied very much. Within allowable limits of these parameters, 'normal' gait can only be obtained if walking speeds are kept around 1.1 m/s to 1.2 m/s.

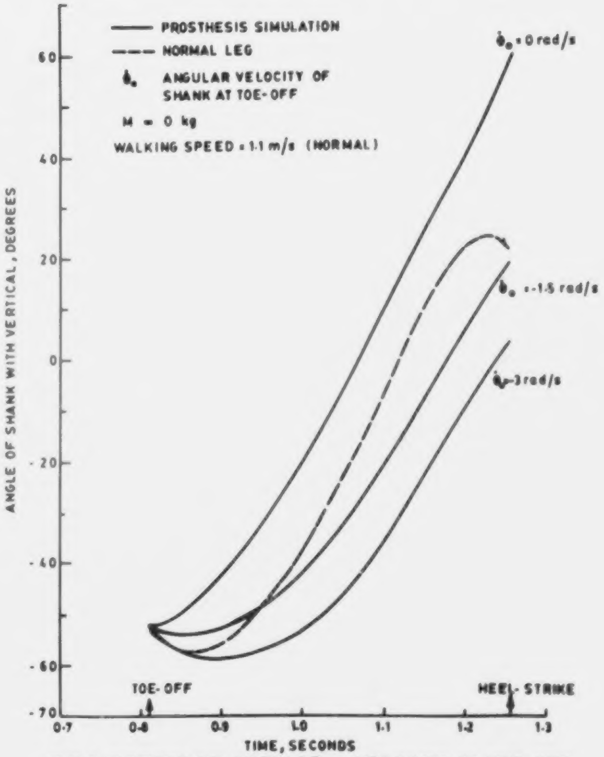

FIG.2 EFFECT OF ANGULAR VELOCITY AT TOE-OFF ON MOTION OF SHANK

CLINICAL OBSERVATIONS

Over 200 such limbs have been followed up for over three years and while problems are being faced with regard to the durability of the knee joints, the amputees seem to be pleased not only because they can squat but because the limb is much lighter. Many amputees who were earlier using the exoskeletal limb were provided with the endoskeletal version and most of them preferred the latter.

The alignment stability provided by the offset knee joint makes them feel so secure that most of them are not required to lock the knee when walking on a level surface. But what strikes one is a remarkably improved swing-phase after a little practice. The stride length and speed of walking is adjusted to an optimum level to provide a near-natural gait. Altered speeds of walking do pose problems. Since there is no friction at the knee, the leg continues to swing till it abruptly comes to a halt with an audible snap. The temptation to add friction etc. was resisted in order to retain a basic simplicity and a low cost. These are important considerations in developing countries.

DISCUSSION

It is slowly being recognized that poor amputees from rural areas in India do not avail of rehabilitation services unless the period of stay at a limb fitting center is of a small duration and the prosthesis provided does not need frequent repairs. The AK prosthesis described above is made from locally available materials and can be fitted in a relatively short time.

The results above indicate that at average walking speeds it is possible to optimize the swing-phase of the shank by an appropriate location of the center of mass and by controlling the timing of the toe-off so that the shank can swing into the right position for heel-strike. The finding that the timing and inclination of the footpiece at the toe-off phase of the prosthesis largely determines the swing-phase characteristic is interesting. Most amputees seem to learn this after a variable period by trial and error, but this observation can be built into the gait training program with considerable advantage.

If the mass of the footpiece is reduced then it would be easier to shift the location of center of mass by shifting of a weight along the shank thus allowing for alterations in walking speed. Therefore it should be feasible to design a simpler alternative to the use of more complicated and therefore more expensive knee mechanisms. Further studies seem to be called for to exploit this finding so that an AK prosthesis more suited for the rural populations of India can be developed.

NOMENCLATURE

g	The acceleration of gravity
I_o	Moment of inertia of the shank and the foot at the center of the knee joint about hinge axis
W	Total mass (shank + foot + variable mass M)
r	Distance of center of gravity (of shank and foot) from the center of the knee joint
y,z	Cartesian coordinates, y-horizontal, z-vertical
Y,Z	Displacements of the knee along y and z directions
θ	The flexion-extension angle of the shank with respect to the vertical, positive in anti-clockwise direction

ACKNOWLEDGEMENT

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STRESS ANALYSIS OF POROUS INGROWTH HIP SURFACE REPLACEMENTS

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ABSTRACT

We examined three new interfaces for porous ingrowth hip surface replacements using linear and nonlinear contact finite element analysis. The results indicate that a flatter interface than is usually used, similar to the epiphyseal line, may improve surface replacement longevity. In addition we find that linear analysis of the pressure at the prosthesis-bone interface gives significantly different results than does an analysis which models the interface as a slip surface. This means that the stress patterns at the interface immediately post-implantation, before bony ingrowth occurs, can not be accurately determined from a simple linear analysis.

INTRODUCTION

It was hoped that total hip surface replacement would be a procedure ideally suited for younger patients. The benefits of the procedure were to be the provision of a functional hip joint with a minimal removal of healthy bone. This would lead to a more natural stress distribution in the post-operation hip and simplify re-operation if it were required. Unfortunately, the failure rates of many types of total hip replacements have been disappointingly high.

Recent research on surface replacement has attempted to incorporate bony ingrowth interfaces on the undersurface of the metal shell. Although promising, this technique has introduced new problems which are associated with bone remodeling in response to altered stress fields.

BACKGROUND

A rigid shell surface replacement destroys the normal pattern of stress transmission in the femoral head. The surface forces are transmitted through the shell to its rim and then in a concentrated manner to the bone at the edge of the implant. The trabecular bone directly under the shell which would carry the load in an intact joint is unloaded or "stress shielded." The addition of a central peg to the shell does not alleviate this stress shielding since the peg tends to transmit the load to the inferior aspect of the femoral neck (1,3).

The stress shielding seen in these designs can lead to a vicious cycle where the stress shielding causes cancellous bone resorption which in turn causes further stress shielding, leading eventually to total failure of the implant (3).

One means of alleviating this problem would be to redesign the prosthesis-bone interface such that the stress is transferred to the bone in a more natural manner (1). A naturally ocurring surface which has potential of being a good interface is the epiphyseal surface of the proximal femur (1,2). Since the epiphyseal plate in young people is composed of cartilage, which is weak in shear relative to cancellous bone, the stress transmission across this surface is due primarily to normal pressure and tension. If the prosthesis-bone interface of a porous ingrowth surface replacement is designed to be similar in shape to the epiphyseal plate a more normal transmission of stress to the bone may result.

MATERIALS AND METHODS

We have examined three new interfaces for the surface replacement: convex, flat, and concave. These interfaces were examined using a 2.54 cm thick 2-dimensional (2D) plane stress finite element model of a transverse slice through the femoral head and neck, (fig.1). The model used bilinear elements. The cancellous bone Young's moduli were 1000Mpa, 500Mpa, 400Mpa, in the head and outer and inner neck respectively. The cancellous Poisson's ratio was, 0.2. The cortical bone Young's modulus was, 5000Mpa, with Poisson's ratio, 0.32. The Young's modulus of the prosthetic implant was variable as described in the following sections with Poisson's ratio, 0.3.

The loading was bilobed, (fig.1), and normal to the articular surface. Each lobe conformed to the equation, $P_{max}\cos^{1/2}\theta$. The vertical resultant was 350.2 Nt/cm. This loading is similar to that expected upon the normal femoral head on the plane of the model.

The prosthesis bone interface was modeled as a perfect slip and as a fixed interface to simulate both the initial implantation and the state of fixation if complete bony ingrowth is achieved. Normal pressure and shear were calculated at the interface but to save space only the normal pressure will be examined here.

RESULTS

Normal pressure at interface

The normal pressure at the interface is important to the rate and degree of bone ingrowth as well as to the degree of bone resorption or deposition at the interface (3). The cases we examine here have Co-Cr prostheses (E_p = 200 GPa) with a convex or concave interface. The pressures are measured in MPa.

Concave Co-Cr prosthesis

The pattern of normal stress under the concave prosthesis at initial implantation, (fig.4), is radically different from the natural stress pattern, (fig.2). The high stresses at the rim are similar to those found for standard shell surface replacements. After bony ingrowth the stresses at the rim are reduced, (fig.3), but the stress pattern is still not similar to the natural pattern.

Convex Co-Cr prosthesis

The pattern of normal stress under the convex prosthesis at initial implantation, (fig.7), is similar to the natural stress pattern, (fig.5), except for the low stresses at the outer rim of the prosthesis. After bony ingrowth the stress pattern, (fig.6), is even more like the natural stress pattern. However, it is questionable whether the bone would ingrow in the rim area since the normal stresses are so low in this area immediately post implantation.

Comparison of interfaces with various materials

The maximum bone stress decrease can be calculated by comparing the stress at a point with the prosthesis implanted, σ_p, with the stress at the same point in the natural femur, σ_n. For three interfaces we plot the maximum stress shielding, σ_p/σ_n, versus the ratio of prosthesis Young's modulus to the cancellous bone modulus, E_p/E_c, with $E_c = 1000$ MPa. σ_p and σ_n are the stress at the point for the prosthetic and normal cases where $\sigma = (\sigma_1^2 + \sigma_2^2 - 2\nu\sigma_1\sigma_2)^{1/2}$, (σ_1, σ_2) = principal stresses. The flat interface is better than the concave and convex cases. Note the rapid onset of stress shielding.

CONCLUSIONS

The epiphyseal scar on this cross-section is flat in the center and downturned at the edges. The excellent stress shielding results of the flat interface imply that a slightly concave interface, similar to the epiphyseal scar, may be an improved interface shape.

The convex prosthesis shows apparently good results. This is largely due to the squeezing action of the load on the femoral head. Since the cancellous bone in the center of the head is replaced by metal it can not be stress shielded as it would be by the concave prosthesis. This again indicates that a prosthesis which has a nearly flat interface may be appropriate.

A distinct disadvantage of a flat design, however, is that it may be unstable when exposed to a wide range of loading directions. Slightly convex or concave designs may therefore offer better initial stability and not incur the stress remodeling problems of current rigid shell designs.

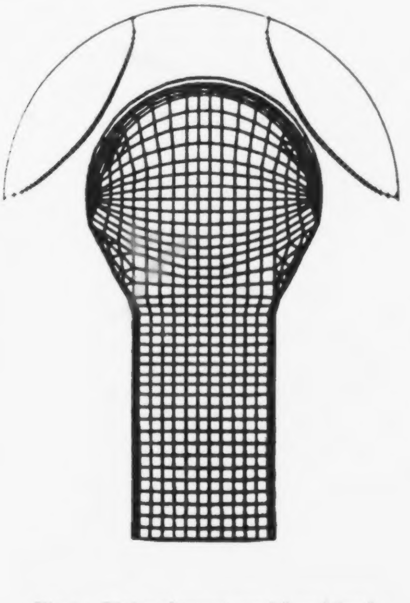

Fig.1 : Finite element model and load.

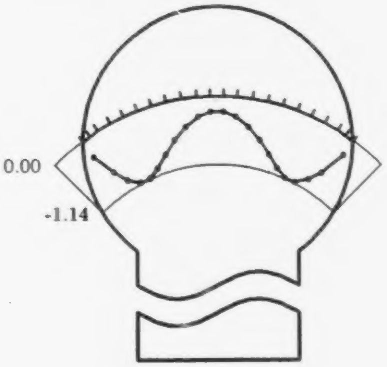

Fig.2 : Normal bone case.

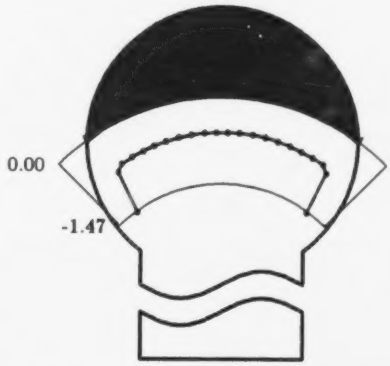

Fig.3 : Co–Cr prosthesis fixed interface.

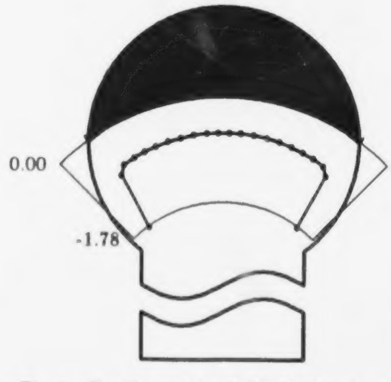

Fig.4 : Co–Cr prosthesis frictionless interface.

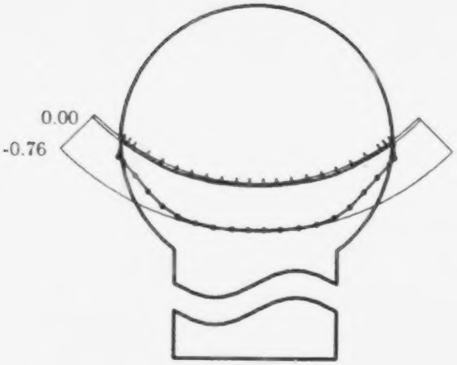

Fig.5 : Normal bone case.

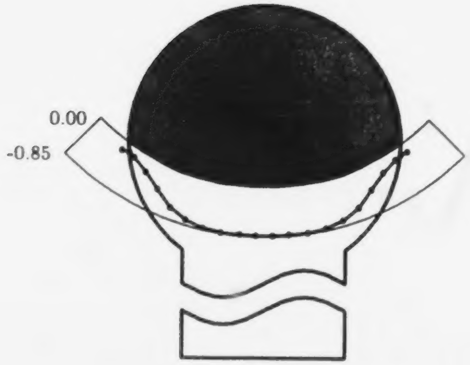

Fig.6 : Co–Cr prosthesis fixed inteface.

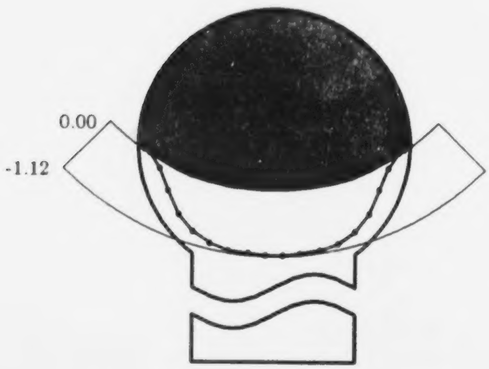

Fig.7 : Co–Cr prosthesis frictionless interface.

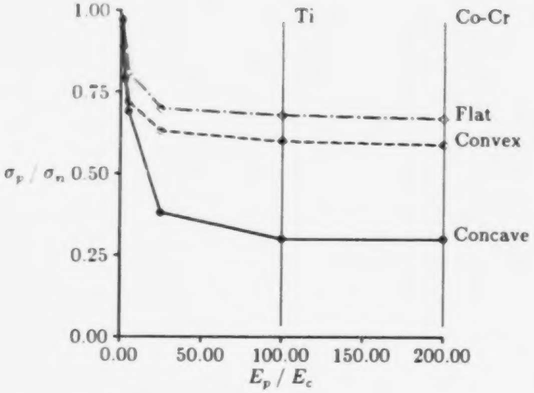

Fig.8 : Maximum stress shielding, fixed interface.

ACKNOWLEDGMENTS

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18.6

FINITE ELEMENT ANALYSES OF POROUS INGROWTH COMPONENTS FOR THE HUMERUS

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ABSTRACT

Finite element stress analyses were performed of the proximal humerus before and after the simulated implantation of stemmed, metallic prosthetic components with porous sintered surfaces for direct bony attachment. Design geometries with surfaces at the prosthetic head/bone interface which were: 1) convex, 2) flat, and 3) concave were studied. Analyses for each of the three geometries were conducted to reflect A) bone ingrowth on all the prosthesis/bone surfaces, and B) bone ingrowth only along the underside of the prosthetic humeral head. Three loading conditions were employed to model various degrees of abduction of the arm. Results indicated that the use of a prosthesis with bone-ingrowth along the stem would cause marked stress shielding proximally while the use of implants with porous ingrowth only on the underside of the humeral head replacement produced stress fields more similar to the normal humerus. The convex, flat, and concave surfaces provided similar load transfer from the component to the underlying bone in all loading cases. Other prosthetic head designs, which may offer better initial stability, produced stress fields similar to those of existing prostheses.

INTRODUCTION

Joint replacement has become a standard procedure in the treatment of patients with arthritis and other crippling diseases. Most of the emphasis in total joint replacement research and development has been centered around the knee and hip components. Total shoulder replacement, by comparison, has received little attention. The purpose of this study was to apply the techniques used in stress analyses of the lower-limbed joints, focussing on the implantation of porous-ingrowth devices.

The successful ingrowth of bone into the implant surface will result in firm fixation and a continuous interface which is efficient in stress transfer. The two important aspects in using these devices are: initial control of the bony ingrowth, and the effects of bone remodeling due to changes in the stress fields. A major challenge associated with the success of porous ingrowth systems is achieving bone ingrowth rather than fibrous tissue ingrowth. It is important to achieve implant stability over a range of loading conditions and to avoid shear and tensile loading at the prosthesis/bone interface. The stress fields in the surrounding bone tissue can be radically changed from those in the normal skeleton upon the implantation of porous-ingrowth components. This has resulted in extreme bone remodeling and eventual implant failure in experimental animal models(1,4,5). Bone remodels to the extent that it can no longer support the implant. The supporting bone fractures and the interface bonding is destroyed. Because of the importance of bone remodeling with the use of porous-ingrowth implants, the questions of where to apply the porous coating have arisen.

In this study, we determined the stress distributions for the normal humerus, and compared them with the stress distributions of implanted shoulder prostheses. We considered three prosthetic designs and examined the difference in stresses between the totally coated porous humeral prosthesis and one with coating only on the undersurface of the humeral head of the implant for each design. The stresses at the prosthesis/bone interface were examined for each design with special attention given to the magnitude of the shear and normal stresses .

MATERIALS AND METHODS

The internal trabecular pattern and geometry were examined and documented with roentgenograms and photographs of the slices. Material properties were assigned to each region based on local bone density and literature values. The models consisted of 2-D planar type elements (constant strain triangles and quadrilaterals) depicting the central coronal section of the humerus (see Figure 1A). The equivalent thickness models were employed to account for out-of-plane stiffness. Each element is considered to be a combination of cortical bone, trabecular bone, and metal (in the analyses with the prosthetic components) as required by their presence and amount within the element region. The equivalent thickness (and, therefore, stiffness) of the region is varied accordingly. In this manner, out-of-plane stiffness can be taken into account in a two-dimensional model. Twenty-four different thickness regions were incorporated into each model.

The finite element mesh was determined from the geometry of the prostheses to be analyzed in conjunction with the designated material properties. Three different implant designs to replace the humeral head were considered. One was geometrically modeled as a "generic" stemmed humeral component with a flat head interface oriented at 45° with respect to the stem (See Figure 1B). The other two designs were similar except that they had curved solid surfaces, one concave (Figure 1C) and one convex (Figure 1D). A total of seven models were analyzed: the normal humerus, three fully porous-coated humeral prostheses and three partially-coated prosthesis, coated only on the underside of the humeral head replacement. To model the characteristics of a uncemented stemmed prosthesis loose in the intramedullary canal, the material properties of the elements representing the stem were changed to trabecular bone for the partially-coated prosthesis analyses. Three different loading conditions were applied to each of the seven models; a 1000 N joint force was applied to the humeral head and a 450 N muscle force provided at the superior facet of the greater tuberosity. Loading case 1 represented the arm at 90° abduction; loading case 2 represented 45° abduction; and loading case 3 represented 0° abduction. Boundary conditions consisted of rigid support of the distal bone.

RESULTS

In the normal humeral head of the equivalent thickness model, the joint resultant compressive forces were transmitted through the trabecular bone to the inferior wall of the humerus. Tensile stresses were created along the superior-lateral region of the bone due to the pull of the rotator cuff muscles. Figure 2 shows the von Mises' strength contours for the normal humerus at 45° abduction. The calculated stress fields correspond with the trabecular bone orientation and density distributions of the cadaver humerus. The different stress distributions of the three loading cases were distinguished by the direction of the principal stresses transmitted to the inferior-medial cortex. In addition, the amount of bending at the distal end increased with each successive loading case, directly reflecting the applied loading at the proximal humerus.

When a fully porous-coated prosthesis of any design was implanted, significant forces were transmitted through the stem of the prosthesis. Stress shielding was created in the proximal humerus and the stress magnitudes in the inferior-medial trabecular bone in the head were greatly reduced (see Figure 3). There was an abrupt increase in magnitude of the bone stresses at the tip of the stem. In the stem and humeral diaphysis, the stresses were higher as the degree of abduction of the arm decreased due to the increase in the bending moment distally. There was little difference in stress distribution among the three design configurations since the presence of the porous ingrowth stem was the dominant feature in determining the stress distributions in all cases.

When the implant was porous-coated only on the underside of the prosthetic humeral head, higher stresses were developed in the inferior-medial region of the humeral head (see Figure 4). Stress distributions were similar to those of the normal humerus. Higher stresses were created in the proximal trabecular bone for loading case 1 than the other two loading cases. The differences in principal stresses and von Mises' strength contours among the three different prosthetic designs were minimal for each loading case.

The stress results were resolved into components in the normal and shear directions at the prosthesis/bone interface. The bone stresses were examined just below the prosthetic head, the most lateral-superior side of the bone represented by zero distance. The normal stresses at the interface were essentially zero or compressive for all three loading cases applied to all designs considered. Because most of the forces in the totally porous-coated prosthesis were transmitted through the stem, little information could be obtained comparing the shear stresses at the prosthesis/bone interface for the three loading cases for each design since all stresses were extremely low. In all the designs with smooth stems where bony-ingrowth occurred on the underside of the prosthetic head, the shear stresses at the interface were lower overall for loading case 2 than the other two loading cases. There was little difference in the magnitude of the normal stresses among the different designs. However, the concave surface prosthesis had higher overall compressive stresses for loading case 2 than the other two designs. In comparing the three designs, no design had lower shear stresses for all loading cases. The flat surface prosthesis had lower shear stresses for loading cases 2 and 3, and the convex surface prosthesis had lower shear stresses for loading case 1. In every case, the normal stresses were higher than the shear stresses.

DISCUSSION

The purpose of this study was to compare the stress distribution of a normal humerus, a totally porous-ingrowth stemmed prosthesis, and a prosthesis with bony-ingrowth only on the underside of the prosthetic humeral head; and to examine different design configurations of the humeral head. The calculated stress patterns of the normal humerus model corresponded to the distribution of density and trabecular orientation observed in the bone tissue of the humeral head. The results indicate that when the humerus is implanted with a totally porous-ingrowth prosthesis, forces will be transmitted down the stem and severe stress shielding will occur in the proximal humerus. The bone stress fields did not differ greatly among the three humeral head designs since the presence of the porous ingrowth stem had a dominating influence. When the humerus is implanted with a smooth stemmed prosthesis with porous-ingrowth only on the underside of the humeral head, the calculated stress distributions are more similar to the normal humerus. Stress shielding in the proximal humerus is avoided to a great extent in all three design configurations for all three loading cases.

As more is being learned about porous-ingrowth technology, one point is very clear: prosthetic design and its influence on bone remodeling is a more important concern with porous-ingrowth devices than with cemented devices. This study emphasizes the observations of other investigators who have reduced the amount of porous-coating on the stem of prostheses. When the stem is firmly fixed inside the intramedullary canal with bony-ingrowth along the stem, the forces are transmitted down the stem distally and the stresses are greatly reduced in the proximal trabecular bone of the humeral head. This will probably result in increased bone deposition distally and bone loss proximally. In animal studies, stress shielding in the proximal femoral head (4,5) and in the distal femur was caused by the implantation of components with stems or pegs (1). In clinical studies, there is evidence of bone hypertrophy at the distal end of the prosthetic stem and calcar resorption proximally (3). In the case of the shoulder, we feel that the implementation of uncemented, smooth stems will reduce deleterious bone remodeling and create stress fields which are more like those of the normal humerus.

The geometry of the implant devices insures that in some locations there will always be load sharing between the implant and bone and therefore, stress distribution anomalies. We examined new designs in order to obtain the maximum benefits from bony-ingrowth fixation. The best design should enable the prosthesis to come in close contact with the bone surface, provide a direct transmission of stress through the implant to the bone tissue, and reduce the amount of shear stresses at the prosthesis/bone interface.

Our results did not demonstrate conclusive evidence that one humeral head design produced better results than the other two. The convex surface appeared to give lower shear stress across the prosthesis/bone interface. If this design were utilized, more bone would have to be removed than with the other designs. The concave surface design created higher normal compressive stresses and shear stresses across the interface. One clear advantage of the concave surface design is that it allows the retention of bone which may be advantageous if revision is required. If the concavity of the head were increased so that a surface replacement shell is created, one would anticipate the same high shear stresses and rim stress concentrations predicted in previous analyses of femoral head shell components (2). For this reason, the use of a humeral shell surface replacement is not advised although a slightly concave head, as modeled in this study, appears to be acceptable. The 45° flat prosthetic head surface appears to provide an adequate surface geometry for porous ingrowth arthroplasty. This geometry approximates the epiphyseal scar in the medial region of the head and facilitates the transfer of compressive stresses across the implant interface while apparently keeping shear stress at an acceptable level. A slightly convex or concave head would probably also be acceptable in terms of interface stress characteristics. Such curved surfaces may additionally provide better initial prosthetic seating and stability than the flat prosthesis during the process of bone ingrowth.

The use of porous ingrowth surfaces on artificial joint components as an alternative to cement is one of the most potentially rewarding areas of research in orthopaedic surgery today. However, close examination should be made of possible adverse bone remodeling due to increase stress or stress shielding with the use of these devices. The use of porous ingrowth should be restricted to surfaces which will best aid in maintaining normal stress distributions rather than using porous-ingrowth on all prosthesis/bone surfaces.

ACKNOWLEDGMENTS

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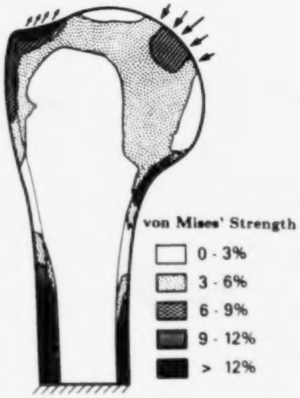

Figure 2

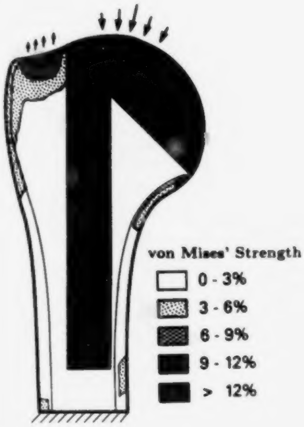

Figure 3

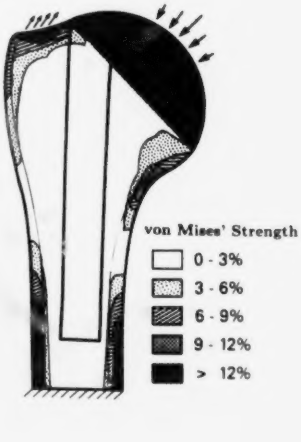

Figure 4

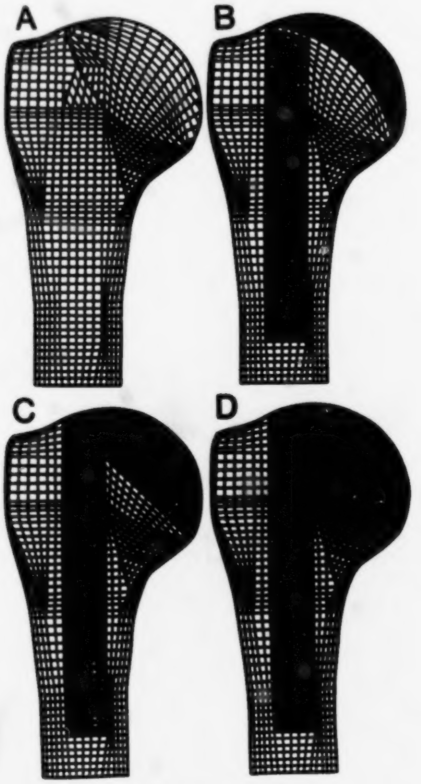

Figure 1

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BONY INGROWTH COMPONENTS FOR THE TIBIAL PLATEAU— A FINITE ELEMENT ANALYSIS

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ABSTRACT

Two dimensional, finite element studies were conducted of the proximal tibia before and after joint arthroplasty. Equivalent-thickness models were created for the natural, proximal tibia and for the proximal tibia with different tibial plateau designs. All components simulated bony ingrowth fixation, *i.e.*, no cement layer existed between the prothetic component and bone. In addition, the interface between component and bone was assumed to be intimately connected, representing complete bony ingrowth and a rigid state of fixation.

Results indicated that conventional plateau designs with central posts or multiple pegs led to higher stresses in the trabecular bone near the distal ends of the post/pegs and stress shielding at more proximal locations. A design without posts or pegs, whose interface geometry mimics the epiphyseal plate minimizes bone stress shielding by providing a loading arrangement wherein the component and bone act in series rather than parallel.

INTRODUCTION

Charnley estimated that in 1976 over 300 different designs for total knee replacements were being marketed. During the same year more than 40,000 total knee replacements (TKR) were performed in the United States alone (1). Since that time, both techniques and designs have improved and one can only assume that significantly more TKR are being performed today. A primary reason for this proliferation of different prosthetic designs has been the lack of long term success due to mechanical problems associated with TKR. Tibial component loosening remains the most frequent complication associated with this procedure. This problem is particularly severe in young, active patients. The use of bone cement as a grouting agent has been implicated as a contributory factor leading to loosening. Recent research has indicated that porous ingrowth fixation may have the potential to decrease loosening rates, however optimal prosthetic designs for such components may be significantly different from conventional designs for cemented prostheses. In view of this consideration, a new anatomic, tibial-component design based on the geometry of the epiphyseal scar was analyzed using two-dimensional finite element models.

The performance of the new designs were evaluated by comparison with the stress fields in the normal tibia and to those created by currently available porous ingrowth components. Both frontal-plane models and sagittal-plane models were created. The frontal-plane models included: 1) the natural tibia; and prosthetic models including: 2,3) conventional designs having a central post and another with short pegs; and 4,5) epiphyseal-based designs with and without a mid-sagittal slit. The sagittal-plane models included: 1) the natural tibia; and prosthetic models including: 2) a conventional design with a peg; and 3) an epiphyseal-based design with an anterior lip. All implant designs had metal backing trays (cobalt chrome) and were modeled to simulate bony ingrowth along the underside of the tray.

MATERIAL AND METHODS

The geometry for the models was obtained from sectioned right and left tibiae of an adult male cadaver with no prior history of joint disease. One tibia was sectioned in the frontal plane, while the other was sectioned in the sagittal plane. After cleaning, these sections were photographed to document trabecular morphology and epiphyseal scar geometry. Estimates of the appropriate material properties of the cortical and trabecular bone were based on literature values (2,3).

The generated models used 2-D elements (constant strain triangles and four node quadrilaterals) of varying thickness. The thickness of each element was established by the principle of equivalent-thickness composite models. Each element was thus considered to be a possible combination of cortical bone, trabecular bone, metal, and polyethylene as required by their presence and amount within the element region. The frontal-plane models had approximately 1500 degrees-of-freedom, while the sagittal-plane models had approximately 700.

The loading applied to the frontal-plane model consisted of: A) two bi-condylar normal forces, (1000 N/condyle); B) a uni-condylar normal force (2000 N); and C) a uni-condylar shear force (1000 N). For the sagittal-plane model, the loading cases analyzed consisted of: A) a 2000 N joint reaction force; and B) a 2000 N joint reaction force acting in conjunction with a patellar ligament force, simulating a particular instant during gait. A total of 16 analyses were conducted with principal stress distributions, failure contours and interface normal and shear stresses being calculated for each analysis.

RESULTS

Frontal-plane model

Shown in Fig. 1 are strength contour plots for the five frontal-plane models subjected to the bi-condylar loading. For the prosthetic designs with either a post or pegs, the stresses are reduced immediately distal to the metal tray and increased near the ends of the post/pegs. In contrast to this, the two epiphyseal designs have strength contours which are similar to those occurring in the natural tibia.

Figure 2 shows the normal stresses at the implant/bone interface for the two epiphyseal designs subjected to the bi-condylar (solid lines) and uni-condylar (dashed lines) loadings. For the uni-condylar loadcase the one-piece design gives rise to tensile stresses under the unloaded condyle. This tipping or tilting phenomenon was also observed with the two models which used conventional prosthetic components and has also been seen *in vitro* (4,5) as well as in previous finite element studies (6,7). The two-piece design, also shown in Fig. 2, did not exhibit this tendency. In this case, the mid-sagittal slit effectively uncouples the motions of the two condyles.

Sagittal-plane model

Figure 3 shows strength contour plots for the three sagittal-plane models. Again it can be seen that the prosthetic design with a peg leads to reduced stresses just beneath the metal tray near the posterior aspect. This is caused by load transmission to more distal sites through the peg. The epiphyseal design provides a series loading arrangement and results in a stress distribution similar to the normal tibia.

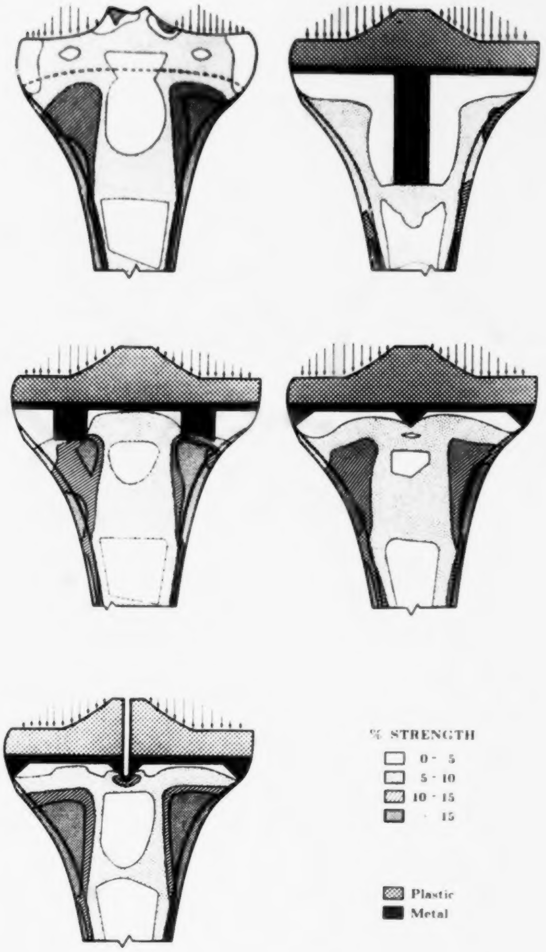

% STRENGTH
0 - 5
5 - 10
10 - 15
15
Plastic
Metal

Figure 1

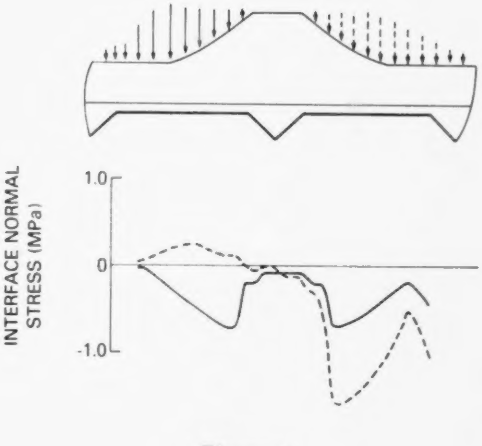


Figure 2a

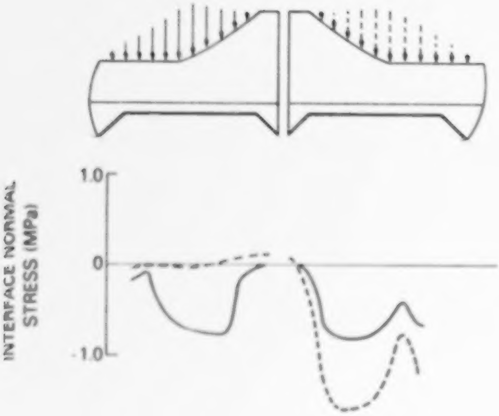
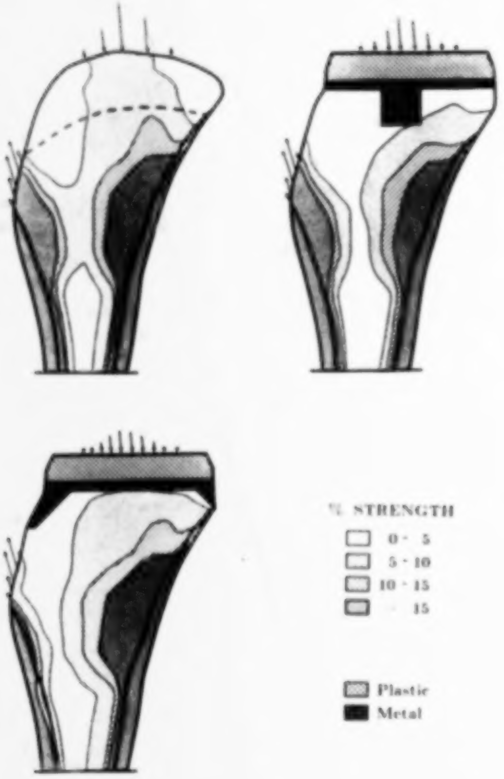


Figure 2b


% STRENGTH
0 - 5
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Plastic
Metal

Figure 3

CONCLUSIONS

Conventional tibial component designs with central posts or fixation pegs significantly alter the stress fields within the supporting trabecular bone, while designs based on the epiphyseal scar geometry result in more benign, near-normal, stress distributions. These designs minimize shear and tensile stresses at the implant/bone interface and also provide near-normal stress distributions, thereby reducing the likelihood of component loosening and excessive bone remodeling. The results of this study, in agreement with previous studies (8,9) indicate that the geometry of the epiphyseal scar can provide a useful guideline for the design of prosthetic components.

ACKNOWLEDGMENTS

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18.8

FEMORAL NECK STRENGTH PREDICTION WITH DUAL ENERGY PROJECTED RADIOGRAPHY

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ABSTRACT

The utility of noninvasive biomechanics is illustrated by our study of the strength of the femoral neck. The objective was noninvasive assessment of the structural strength of this clinically important region. Femur specimens were subjected to dual energy digital radiographic scan in a cylindrical phantom filled with saline which simulated soft tissue. Soft tissue subtracted images obtainable by the dual energy scan procedure were analyzed by a procedure duplicating that which could be performed on a patient. The femora were then loaded in a standard servohydraulic materials testing system until fracture occurred at the femoral neck. A model of the femur as a nonhomogeneous beam of arbitrary cross section subjected to combined bending and axial loading was fitted to the experimental force data. A multiple correlation coefficient of 93 percent was obtained. This result was compared with those for several other strength predictors current in the literature and found superior to them. This indicates that proper biomechanical analysis can provide bone strength estimates superior to those obtainable from intuitive analyses of images.

INTRODUCTION

Noninvasive imaging offers new vistas for biomechanics research and for clinical applications of biomechanics. Much work formerly feasible only in specimen studies on cadaver material can be done on living subjects using increasingly available digital imaging techniques. Such techniques include digital radiography, computed tomography and magnetic resonance imaging among others. Generally speaking, the visible structure contours on the image provide information about the geometry of body organ structures while the intensity of points on the image is indicative of material characteristics such as density and average atomic number. Appropriate calibration allows values from the image to replace those obtainable from a specimen for application in such areas as beam theory modeling, finite element modeling, and kinematic study. Noninvasive imaging, thus, opens up possibilities such as stress analysis of a patient's unique individual body structures or kinematic analysis of an individual's own body movement possibilities. While these applications are important to society at large, they are particularly important in the rehabilitation community where patients frequently present unusual orthopedic configurations.

Aging, immobilization, and weightlessness of astronauts in space are among factors which have been linked with bone loss. A possible consequence is structural inadequacy for the activities of daily living. Conditions of osteopenia and osteoporosis in postmenopausal women and in the elderly of both sexes can be expected to pose an increasing significant social problem as our population ages. Hip fracture constitutes, perhaps, the most serious consequence in terms of morbidity and mortality. It has been estimated that over 200,000 hip fractures occur annually in the U.S., that the associated annual cost of patient care exceeds 750 million dollars, and that 19 percent of patients die as a result of the fracture (1). Determination of femoral neck strength is, then, a worthy goal for application of noninvasive biomechanics.

BACKGROUND

Controversy has surrounded the proper measurement site for bone mass. A determination of osteoporosis is frequently made via iliac crest biopsy or measurement of bone mineral content at the distal forearm. It has been shown, however, that the bone mass in one body region may bear little relationship to the bone mass in another (2). For this reason it seems best to attempt measurement of femoral neck weakness by assessment of the femoral neck region itself. Noninvasive digital imaging of the femoral neck followed by application of biomechanical principles to yield strength predictions provides a logical procedure for obtaining regional strength measurements.

Many previous attempts at noninvasive bone strength assessment have focused on quantification of bone density or bone mineral content while ignoring the simultaneously important structural geometry. Other studies, in contrast, have measured geometric variables such as cortical thickness while ignoring material properties such as density. Since structural strength is governed by the interaction of geometry and material properties, unified consideration of these factors is necessary for accurate prediction of bone organ strength. Explicit mechanical models of structural performance were similarly hypothesized more effective than intuitively conceived bone densitometric variables as predictors of bone strength.

Only a few biomechanically based analyses of femoral neck strength have been reported. Strong assumptions about femoral neck geometry characterize these pioneering studies (3,4). In the analytical procedure to be outlined here trigonometric correctness is automatic, no assumptions are made regarding cross sectional geometry, and a simple algebraic relationship accurately predicting femoral neck strength for the experimental loading condition is obtained.

METHODS

Eleven embalmed human femora were subjected to dual energy scanned projection x-ray imaging (5). Placement of each femur in a saline-filled lucite cylinder during the scan simulated the presence of soft tissue. After scanning, each femur was loaded to failure by applying a simulated joint force directly to the femoral head in a servohydraulic materials testing machine. A universal joint in the lower grip of the test fixture simulated the knee joint and protected the materials testing system from damaging bending moments.

The soft tissue subtracted image available from the dual energy imaging procedure was analyzed by placing lines on the femur images to indicate cross-sections of interest and plotting the image intensity profiles for each line (Figure 1). The zero baseline for each image intensity profile in the figure is a line parallel to the cross sectional line but below it and to the left. The concept is illustrated in Figure 2 which also shows the mathematical interpretation of the profile. Precalibration of a phantom placed in the field of view with the femur allowed expression of the intensity of each pixel (i.e. point) in a profile as an equivalent area density of bone (g/cm^2). The profiles thus may be viewed as density weighted thickness profiles. Each pixel value represents the line integral of bone apparent density over the thickness of bone orthogonal to the plane of the image.

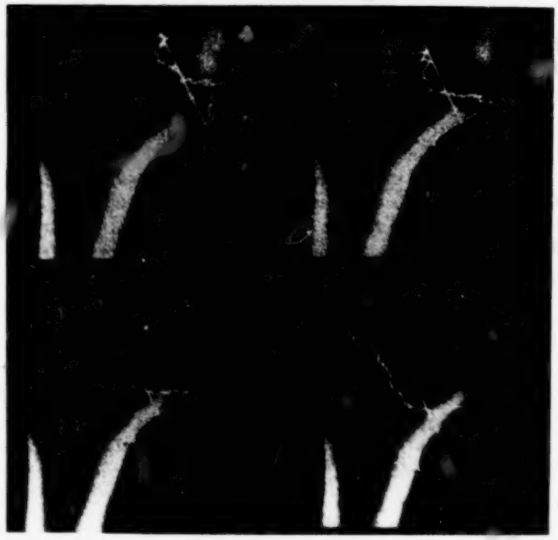

Figure 1. Image analysis

Integrals of this density weighted thickness profile over the bone width (6) yielded the density weighted area, the center of mass, and the density weighted moment of inertia for each cross section:

Density weighted area	$\hat{A} = \int_0^w g(x)\,dx$
Center of mass	$\bar{x} = \frac{1}{\hat{A}} \int_0^w x\,g(x)\,dx$
Density weighted moment of inertia	$\hat{I} = \int_0^w x^2 g(x)\,dx - \bar{x}^2 \hat{A}$

The cross-sectional properties and width of each section were combined into appropriate terms for predicting bone strength according to various mechanical models.

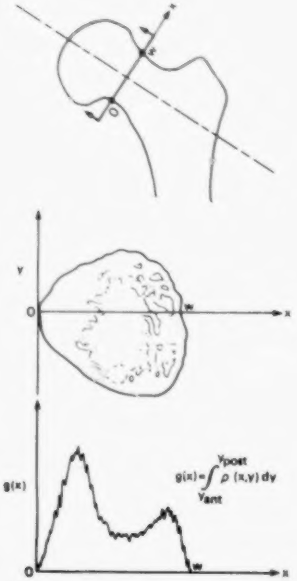

Figure 2. Intensity profile interpretation

The most general mechanical model used envisoned the femoral neck as a nonhomogeneous beam of arbitrary cross section subjected to a loading causing both axial compression and bending (Figure 3). The longitudinal stress is the sum of the normal and bending contributions respectively

$$\sigma = \frac{-(F\cos\alpha)}{A} + \frac{(F\sin\alpha)L(w-\bar{x})}{I}.$$

This may be rearranged and written for nonhomogeneous specimens as

$$\frac{1}{F} \approx \frac{\bar{\rho}}{\sigma}\left[\frac{-\cos\alpha}{\hat{A}} + \frac{L(w-\bar{x})\sin\alpha}{\hat{I}}\right]$$

which can also be expressed as

$$\left(\frac{1}{F}\right) \approx a\left(\frac{1}{\hat{A}}\right) + b\left(\frac{w(w-\bar{x})}{\hat{I}}\right)$$

where a and b are parameters to be estimated. This model was fitted to the inverse of the experimental failure loads by multiple regression (multivariate least squares).

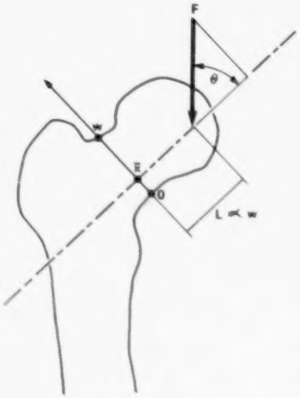

Figure 3. Coordinate system

RESULTS

To compare the strength predictions emerging from our nonhomogeneous beam theory analysis with other indicators of bone condition current in the literature we plotted the failure load for the experimental femora against various strength indicators which have been suggested. The relationship of bone strength with age has often been emphasized. For our femora a trend toward lower load bearing capability with age was noted but the scatter was large.

The density weighted cross sectional area obtained from our images corresponds to the bone mineral content obtained by single or dual photon absorptiometry, e.g. using the Norland-Cameron bone mineral analyzer. At the cross section providing highest correlation, i.e. the subcapital cross section, correlation between density weighted area and failure load was roughly 80 percent when a constant term was permitted in the regression equation. Allowing this term means that the failure load and density need not become zero at the same time. Disallowing this term to force satisfaction of the more realistic condition that zero density and zero load bearing capability must correspond, lower correlation was observed.

To assess the relationship of failure load to the average apparent density of the combined cortical and cancellous bone at the femoral neck, an elliptical cross section was assumed. The ratio of the major to minor axes was set by measurement of one specimen. For each cross section the ellipse size was set by the width of that cross section on the image. Division of each density weighted area measurement by the corresponding elliptical area estimate provided the average density of the bone in the cross section. The correlation was lower than that for the relationship between failure load and density weighted area but was again highest at the subcapital cross section.

The relationship of the actual experimental failure load to the failure load predicted by the nonhomogeneous beam theory model showed a strong multiple correlation coefficient at all cross sections analyzed. Strongest correlation was at the most distal cross section where the multiple correlation coefficient was 93 percent when a constant term was incorporated in the regression equation. The relationship is plotted in Figure 4. As mentioned above, a weaker relationship was obtained for pure compressive models. A still weaker one was obtained for pure bending models. This indicates the presence of both axial loading and bending in the loading configuration employed experimentally.

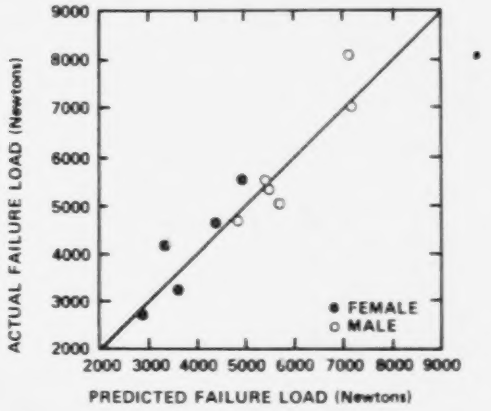

Figure 4. Nonhomogeneous beam theory model

DISCUSSION

The better performance of density weighted area than of density itself as a predictor of femoral neck failure load emphasizes the simultaneous importance of material properties and structural geometry as determinants of the strength of a structure. This means that any analysis which investigates either bone density or bone geometry while ignoring the other will be incomplete.

The better performance of the nonhomogeneous beam theory model than of the density weighted area as a predictor of the failure load is to be expected. Note that if the bending term is neglected in the nonhomogeneous beam theory model equation, a proportionality relationship between the failure load and the density weighted area is obtained. This means that a relationship between failure load and density weighted area is the nonhomogeneous beam theory model for a case in which no bending component of load is present. Were the femur loaded in pure axial compression along the neck, the density weighted area measurement would be the correct predictor of its strength. While physiological loadings on the proximal femur need not correspond to the simple loading condition used in our experimental work, it is anticipated that most physiological loadings will involve both bending and axial loading making a "bone mineral content" measurement alone inadequate as a strength predictor.

The success of the nonhomogeneous beam theory model for prediction of failure strength for the experimental loading condition suggests possible clinical utility for such a strength procedure. By assuming the fitted regression equation to hold, the failure force corresponding to the combination of sectional properties obtained from analysis of a patient scan can be estimated. Effective clinical strength prediction by this method will result if bone loss from the osteoporotic femoral neck dominates loading conditions as the causative agent of fracture.

In this study dual energy projection digital radiography allowed relatively effective prediction of experimental femoral neck strengths. Combination of biomechanical analysis with other imaging modalities should be similarly useful. Computed tomography, for example, could be used to model biaxial bending and torsional effects in the femoral neck ignored in this study. By basing finite element models rather than nonhomogeneous beam theory models on the output of a CT scanner, bone structures involving geometries unlike beams can be modeled. Recognition of the biomechanical modeling assumptions implicit in quantitative computed tomography, a density weighted area measurement applied to vertebral bodies, could suggest new directions for spinal osteoporosis research. The possibilities opened by noninvasive imaging as an avenue for biomechanical analysis of an individual's unique structural features are nearly limitless. For amputees the structural adequacy of the bone in a stump could be quantified indicating its ability to support the patient with a prosthesis. Analysis of remodeled bone around an implanted prosthesis could indicate whether prosthesis loosening is imminent. Development of appropriate procedures for performance of biomechanical analyses based on noninvasive images, thus, presents an important challenge for advancement of health care.

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The Mechanical Properties of the Lateral-Collateral Ligaments of the Human Ankle Joint

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ABSTRACT

In this study the mechanical properties of the Lateral-Collateral ligaments of the human ankle joint were investigated. Tensile force-elongation testing at slow strain rates were performed on bone-ligament-bone preparations. From these tests the stiffness constants, ultimate loads, elongation to ultimate load and mode of failure were obtained.

The results indicate that the Anterior Talo-Fibular ligament has the higher stiffness values with failure occuring at the smallest elongation. In addition, the Anterior and Posterior Talo-Fibular ligaments demonstrated higher ultimate loads than the Fibulo-Calcaneal ligament. The results suggest that the high frequency of injury to the Anterior Talo-Fibular ligament is strongly related to the location of the ligament relative to the joint rather than to mechanical inferiority. In future studies the tensile stress-strain characteristics of these ligaments will be investigated.

INTRODUCTION

The ankle sprain is probably one of the most common injuries which the physician is called upon to treat. Of all ankle sprains, the lateral ligaments are most frequently injured. Isolated tear of the anterior talo-fibular ligaments comprises 65 - 80 percent of all treated cases (1,2). Primary treatment of most ankle sprains, consists of conservative non-operative methods. However, for the more severe sprains involving complete rupture of ligaments, surgical intervention may be required. Such surgical intervention often requires the replacement of the damaged ligament by a tendon (tendonesis) or by an artificial ligament (3,4,5,6).

For successful repair of severely injured lateral ligaments any natural or artificial replacements must have mechanical properties similar to the original ligaments. However, there seems to be a lack of data describing the mechanical properties of the ligaments of the human ankle joint. The purpose of this investigation is to provide this information. In addition, data on the mechanical properties of the ankle ligaments can be used to study, through an analytical model, the overall stiffness characteristics of the ankle joint, and the contribution of the individual ligament to the stability of the joint.

EXPERIMENTAL PROCEDURE

The experiments were performed on five embalmed lower limbs. Following the removal of soft tissue surrounding the ligaments, a sagittal cut was made through the ankle so as to isolate the lateral from the medial ligaments. The testing was performed on bone-ligament-bone preparations since securing of a firm grip of the ligament into the testing apparatus was extremely difficult due to the short length of the ligament. Ligaments were separated from the corresponding bones leaving a small bony attachment around their area of insertion (Fig. 1). The bony attachments of each ligament were embedded in PMMA and a 3/8 in. diameter pin was inserted into each of the bone-PMMA units. The specimen was then mounted with the use of specially designed grips into a tensile testing machine (Fig. 1). By using ball and socket joints at each end it was ensured that the applied force was aligned along the direction of most of the ligament fibers.

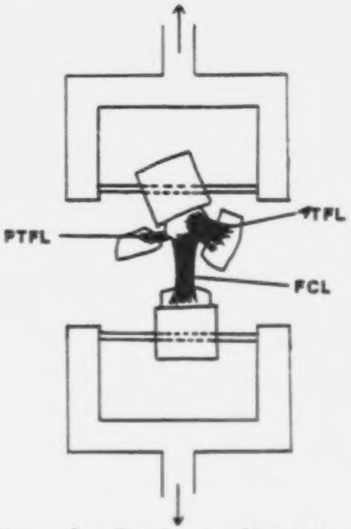

Figure 1. Testing configuration of bone-ligament-bone specimen

Following these preparations, a tensile test at slow strain rate (.125 in/min) was conducted. In this test, the ligament was initially subjected to ten pre-conditioning loading cycles of approximately 30 - 50 percent of ultimate load, then loaded to 50 - 70 percent of ultimate load for one cycle, and then brought to failure. During the test, the applied load and the grip to grip displacement were continuously recorded. The mode of failure of each specimen (avulsion or tear) was observed and registered.

DATA PROCESSING AND RESULTS

Figure 2 represents a typical example of load-elongation curves for the lateral collateral ligaments of one ankle joints.

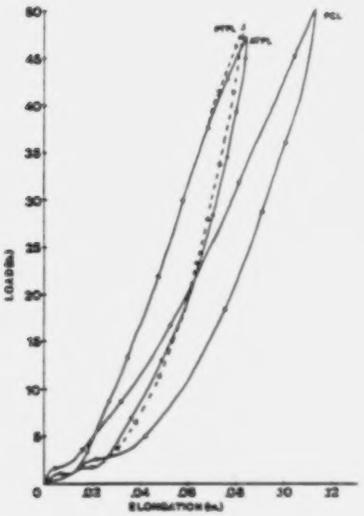

Figure 2. Example of load-elongation characteristics for the lateral-collateral ligaments.

ATFL -Anterior Talo-Fibular ligament
FCL -Fibulo-Calcameal ligament
PTFL -Posterior Talo-Fibular ligament

Preliminary tests indicated that the first few loading cycles provided different results than subsequent tests. The ten pre-conditioning cycles reduced significantly the subsequent cycle to cycle variability (see also 7,8).

The results indicated a load-deformation behavior typical of soft tissues, with stiffness increasing with elongation and an apparent linear region at higher deformations. The force-elongation relationships obtained were approximated by linear increments. From there, the secant stiffness constants K_i, $i = 1,5$ were obtained (Fig. 3). Additionally standardized zero reference load (L_0) and elongation (E_0) values were calculated.

Observations indicate that the modes of failure were ligament tears, a proximal or distal avulsion, and in one case fracture of bone. Mode of failure for each ligament as well as K values, ultimate load, and elongation to ultimate load values are shown in Table 1.

DISCUSSIONS

Comparison of the load-elongation characteristics of the three lateral collateral ligaments indicate that: a) the anterior talo-fibular ligament generally demonstrated the highest stiffness values, b) the anterior and posterior talo-fibular ligaments demonstrated higher ultimate loads than the fibulo-calcaneal ligament, c) the anterior talo-fibular ligament failed at

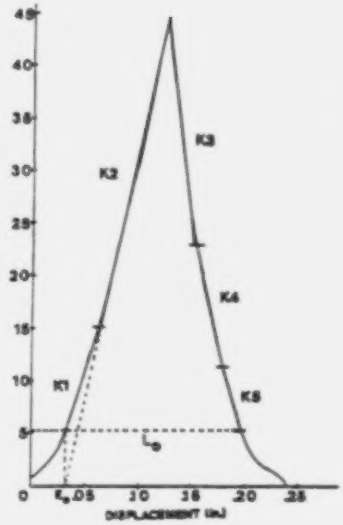

Figure 3. Incremental linear approximation of load-elongation properties of ligament.

K_i - stiffness constant,
E_0 - reference elongation
L_0 - reference load

LIG.	K1 lbs/in	K2 lbs/in	K3 lbs/in	K4 lbs/in	K5 lbs/in	ULT. LOAD lbs	ELONG. TO ULT. LD. in	MODE OF FAIL. (TYPE)
ATFL I	540	[illegible]	-1533	-766	-382	46	.08	DA
FCL I	276	513	-1111	-462	-318	79.5	.13	DA
ATFL II	[illegible]	1348	-2320	-1389	-720	123.5	.10	PLT
FCL II	320	480	-840	-580	-280	76.5	.18	PLT
PTFL II	440	700	-1290	-700	-480	80	.16	MLT
ATFL III	[illegible]	[illegible]	-1278	-773	-420	52.5	.07	DA
FCL III	300	580	-1260	-440	-380	67	.20	MLT
PTFL III	440	730	-1400	-480	-360	75**	.11	-----
ATFL IV	[illegible]	[illegible]	-1300	[illegible]	-420	73.5	.13	DA
FCL IV						17*		DA
PTFL IV	525	920	-1520	-760	-380	150	.17	PA
ATFL V						28*		DA
FCL V	376	560	-780	-580	-320	64	.23	PA

Table 1. Results obtained from tensil force-elongation tests.

DA = distal avulsion
PA = proximal avulsion
MLT = mid-ligament tear
PLT = proximal ligament tear
DLT = distal ligament tear

* specimen failed in pre-conditioning cycle
** bone fracture.

the shortest elongation. These results suggest that the high frequency of injury to the anterior talo-fibular ligament is related to the location of the ligament relative to the ankle joint rather than to mechanical inferiority. Examination of Table 1 indicates that the most common mode of failure was avulsion. This supports Noyes et al. observation that avulsions predominate at slow station rates (10).

There appear to be large inter-specimen variations in the mechanical properties of the ligaments. These variations may be due to storage conditions, donor's age, and ligament size (8). Therefore, future studies will involve the use of fresh specimens. These specimens will be grouped according to donor age, and in addition to load-elongation values, stress-strain characteristics will be measured at varying strain rates.

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19.1

INTELLIGIBILITY AND MACHINE RECOGNITION OF DEAF SPEECH

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ABSTRACT

SRI used deaf speakers to test the hypothesis that a person whose speech is unintelligible to human listeners, may be consistent and distinct enough so that an automatic speech recognizer can recognize the speech. We tested this hypothesis by comparing the recognition accuracy of a high-performance, template-based speech recognizer to that of human listeners. A limited vocabulary of 40 words spoken by five vocally-impaired deaf talkers was used. Comparing intelligibility to recognition accuracy, three of the five deaf speakers were more than 60% better understood by the recognition device than by human listeners. Thus, these three speakers might be good candidates for a voice-in/voice-out communication aid.

INTRODUCTION

At the ICRE II (RESNA 1984) meeting in Ottawa, Stevens et al (1) described experiments that validate an approach to deaf-hearing dialogues over the telephone This approach uses speech recognition in the hearing-to-deaf direction and, if the deaf person's speech is unintelligible, uses speech synthesis in the deaf-to-hearing direction For a vocally-impaired deaf person, one issue in that research was: how can the person most efficiently enter text into a synthesizer to generate speech at conversational rates? Standard typing is slow (deaf subjects typed in the 10 to 50 word/minute range); an anticipating typewriter only yielded small rate increases (compare (2)); and the *vocally impaired* deaf subjects had great difficulty with the phonetic representations required to learn stenotype.

Our research at SRI (3) suggests that some deaf people may be able to speak (in isolated words) faster then they can type, so that if their speech maintains consistent and distinct forms of a sufficient number of words, the deaf-to-hearing direction of the telephone link might be accomplished by voice-in/voice-out technology.

There has been substantial work on communication aids for vocally impaired people (4), much of which has focused on the problem of message generation by the user and on the methods for entering the message into the aids for consequent display, printing or voice output (e g 5,6,7,8). Communication aid research has generally focused on message input because text and voice output are well supported commercial technologies.

Previous Voice Input Communication Aids

The available data on the potential of voice input for communication aids has mostly consisted of attempts to construct such devices for particular speech impaired individuals. Margaret Barker of Children's Hospital at Stanford evaluated a recognition input device with a dysarthric subject and reported the "speech recognition accuracy was only 40%, and many words had to be repeated five to seven times before response"(9). Mclanie Fried-Oken (10) at Tufts New England Medical Center tried a commercial speech recognition device with two patients, a 10 year-old respirator-bound quadraplegic and a 19 year-old paraplegic with flaccid dysarthria. She acheived no success with the respirator-bound patient, but reports over 87% accuracy using the recognizer with the dysarthric.

In a recent Phd. thesis project at the University of Sourthern California, Wissam Ahmed (11) and his advisor, George Bekey, studied articulatory and temporal aspects of five dysarthric cerebral palsy speakers. One cerebral palsy speaker was chosen for testing with an automatic speech recognition system developed at USC. The speaker read a list of 20 words five times, and speech recognition experiments were performed in which the recognition algorithm, the number of training passes and the signal processing parameters were modified. Recognition rates for the cerebral palsy speech ranged from 53% to 87%, depending on the algorithm and parameters used.

METHOD

SRI used deaf speakers to test the hypothesis that an impaired speaker whose speech may be unintelligible to human listeners, may be consistent and distinct enough so that an automatic speech recognizer can recognize the speech. We tested this hypothesis by comparing the recognition accuracy of a high-performance, template-based speech recognizer to that of human listeners. A limited vocabulary of 40 words spoken by five vocally-impaired deaf talkers was used. Machine and human recognition of the speech of normal hearing talkers were studied as a control.

Subjects

We recorded 11 subjects, five deaf and six hearing. The hearing subjects (four females and two males ranging in age from 22 to 38) all had normal speech and none reported any hearing loss. The deaf subjects (four females and one male ranging in age from 26 to 35) all had prelingual severe to profound hearing losses and had a definite "deaf" speech quality. All five of the deaf subjects had some oral training when they were young, but only two currently use speech in their daily communication. All five deaf subjects use American Sign Language, instead of spoken English, as their primary mode of communication.

Materials

Subjects read from a non-diagnostic articulation test designed by Templin (12). The list is comprised of 40 common words that deaf readers are likely to know, and it includes many consonant clusters and several easily confusable word pairs (eg. grass/glass, swing/spring, etc.). The Templin list is:

reaching
skate
brother
shoe
splashing
snow
yellow
cherry
sled
anything
grass
cloth
there
flower
spring
thumb
truck
valentine
spoon
jump
throw
present
shrink
swing
drum
stove
glass
scratch
fresh
smooth
cracker
zipper
pleasure
larger
onion
bird
straw
music
twins
queen

Recording

Each subject was recorded on two separate days, the second recording being one week to eleven days after the first. On each recording day, six different randomized orders of the above 40-word list were recorded onto audio tape using a Nagra IV-S tape recorder and a Bruel & Kjaer 4132 microphone. Thus, a total of 12 recordings of each word were obtained for each subject in two recording sessions. The subject was prompted to speak one word at a time by displaying each word on a CRT for two seconds with a one second pause between words. The recording sessions included a short break between the second and third readings and between the fourth and fifth readings of the list. If subjects stumbled on any word, they were immediately asked to repeat the word, ensuring that 12 good versions of each word were available for each subject.

Automatic Recognition Testing

After all twelve lists were recorded, the words were processed by a template-based speech recognizer developed at U.C. Berkeley (13). For each speaker, each of the 480 templates (40 words spoken 12 times) was automatically compared with the other 479 templates by the recognition device, and distance scores between all templates were generated. The scores for the templates were ranked in order of similarity, and the distributions were analyzed.

Listener Testing

We tested four of the deaf speakers, ranging from the least consistent to the second most consistent. For each of the deaf speakers, two of the twelve original recordings were selected at random to be played to hearing listeners from a Sony TCM-5000 tape player over a small loudspeaker in a quiet room. The listeners were unfamiliar with the vocabulary and the speakers. The order of presentation of the two lists from a speaker was balanced across listeners, in case one rendition was more intelligible than the other. For each deaf speaker, six hearing subjects listened to the two recordings.

First, the hearing subject heard one of the twelve recordings of the 40-word list spoken by the deaf speaker, one word at a time, with no information about the vocabulary, and the listeners were instructed to write each word as they heard it (open set). Listeners were allowed to request that a word be repeated, in which case the tape was rewound and the word was played again. In this open set test, pure intelligibility of each word was measured with no contextual information.

Next, the same hearing subject heard the same words spoken again in a different order by the same deaf speaker, but this time the hearing subjects were shown the list of words and told to circle the word they heard (closed set). Again, they were allowed to have a word repeated if they wanted. The listeners responded to each word on a new response form that had the 40 words printed in a different order, so that only one word was circled on each form. This reduced the likelihood that listeners chose words by elimination.

RESULTS

Table 1 shows the percent of templates that were correctly recognized, i.e. cases in which the closest template (the recognizer's first choice) was another token of the same word. If the closest template was a token of a different word, the recognition was counted as incorrect. Table 2 shows the percent correct for human listeners.

The automatic speech recognizer recognized the speech of the deaf speakers better than the human listeners. For example, speaker CR, the least intelligible of the deaf speakers, had about 75% or her words correctly identified by the speech recognizer, whereas human listeners understood only about 7% of the words as an open set task and 30% as a closed set. 75% recognition by machine (even of a limited vocabulary) may be enough to allow user CR to communicate by voice with the hearing population; she is now unable to be understood when she uses her voice. About 80% of speaker MR's words were correctly recognized by the device, but only 9% were recognized by listeners. For speaker SS, the machine identified 91% of her words, whereas the listeners understood 21%. For deaf speaker DH, the machine recognized 98% of his words, the listeners identified 61%.

Comparing intelligibility to recognition accuracy, three of the five deaf speakers were more than 60% better understood by the recognition device than by human listeners. Thus, these three speakers might be good candidates for a voice-in/voice-out communication aid.

Table 1

Percent Correct Word Recognition by an Speaker-Trained Automatic Speech Recognizer with a 40 Word Vocabulary

Deaf		Hearing	
subject	% correct	subject	% correct
RF	99.2	RH	100.0
DH	97.5	EG	100.0
SS	90.8	OK	100.0
MR	79.6	KE	99.8
CR	74.6	DW	99.6
		ES	98.5

Table 2

Percent correct word identification in open set and closed set responding by naive Human Listeners

Listener	Deaf Speaker	Open Set % correct	Closed Set % correct
1	DH	62.5	97.5
2	DH	55.0	92.5
3	DH	65.0	90.0
4	DH	75.0	87.5
5	DH	62.5	90.0
6	DH	45.0	92.5
average	DH	60.8	91.6
7	SS	17.5	57.5
8	SS	22.5	52.5
9	SS	17.5	52.5
10	SS	22.5	67.5
11	SS	15.0	65.0
12	SS	30.0	65.0
average	SS	20.8	60.0
13	MR	7.5	60.0
14	MR	10.0	45.0
15	MR	12.5	50.0
16	MR	5.0	50.0
17	MR	10.0	57.5
18	MR	10.0	67.5
average	MR	9.2	55.0
19	CR	5.0	27.5
20	CR	5.0	17.5
21	CR	7.5	32.5
22	CR	10.0	42.5
23	CR	10.0	27.5
24	CR	5.0	32.5
average	CR	7.1	30.0

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A MICRO-PROCESSOR AND SIGNAL PROCESSOR BASED SPEECH TRAINING AID FOR THE HEARING-IMPAIRED

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ABSTRACT

The shortage of highly trained speech therapists has created a need for a speech training system which a trainee can use, independently or with a therapist, through a number of different stages of development which differ markedly in the goals which must be attained. A relatively simple and inexpensive micro processor and signal-processor based speech training system is designed which is able to be appropriately utilized in many of these stages, including basic vocalization, intensity, duration, and pitch control, and the pronounciation of both vowels and dipthongs. The system is designed to minimize overall costs by reducing hardware requirements.

INTRODUCTION

The shortage of highly trained speech therapists has created a need for a speech training system which the therapist can use to enable a hearing-impaired person to engage in effective speech training independently. An inexpensive and effective general-purpose training aid would fill a very definite need in helping to alleviate some of the demands on the speech therapist's time.

For persons who became deaf before the age of five, the acquisition of intelligible speech is an almost overwhelmingly difficult task. To succeed in this effort requires a vast amount of time and highly individualized instruction through a number of different stages of development which differ markedly in the goals which must be attained. Clearly defining these goals helps greatly to facilitate progress through these stages and helps both student and teacher to make the most effective use of their time [1]. A training system which is designed to be able to be appropriately utilized in many of these stages could be an enormous asset in the teaching environment. This system is designed in order to be able to be utilized at each of the following three stages : first, the development of basic vocalization; second, the ability to control variation in intensity, duration, and pitch; and third, the production of specific vowels and dipthongs. Since the system has the capability of generating real-time displays of both pitch and intensity, it can also be used in later stages for training in the production of speech with natural voice patterns.

MATERIALS AND METHODS

The System

The speech training system is a relatively simple and inexpensive micro-processor and signal-processor based device. The hardware of the system (Z-80 processor, four NEC uPD7720 signal processors, low-pass filter, A/D convertor, and CRT display) has been described in a previous paper [2], but all of the software and the training methods implemented have been completely re-worked.

Vowel Analysis. In the analysis of the speech input during vowel training, two of the four signal processors are used as before to compute the linear predictive coefficients of a 25.6 msec length sample of the signal. However, in its current configuration, both of the remaining signal processors are used in computing the spectral envelope of the signal from the LPC coefficients, providing reliable estimates of the spectrum at 50 Hz intervals from 0 to 3850 Hz. The signal processors locate all of the potential peaks in the resultant spectrum through an analysis of its first and second difference. The locations of these peaks are returned to the main processor which uses a simple heuristic method to identify potential candidates for the first and second formants. Since the difference limens of the human ear in distinguishing formant frequencies has been estimated to be in the range of 20-60 Hz [3], this should be sufficient accuracy. The time required to input, analyze, and display the formants extracted from one 25.6 msec frame is 58.8 msec, enabling the system to provide feedback to the user about 17 times per second.

Pitch Extractor. In designing a system to provide speech training for the hearing impaired, it is essential to be able to extract the fundamental frequency, or pitch, of the trainee's input voice signal. One of the major difficulties that the hearing-impaired encounter in attempting to learn to speak is learning to control the pitch of their utterances in order to produce intelligible, natural sounding speech.

An almost bewildering variety of pitch extraction techniques have already been developed due to the importance of pitch detection in many different speech-processing systems. With such a wide variety of algorithms available, one would expect to be able to find one that would be appropriate for almost any given application. In the present case, there were two basic requirements which had to be met by the pitch detection algorithm used. The first was that the algorithm must operate in real-time, or at least close to real-time, in order to be able to generate continuous pitch displays to be used in basic vocalization and intonation training. The second requirement was to avoid increasing the cost of the system by using an algorithm which necessitated the inclusion of extra hardware, such as an additional low-pass filter or signal processor. The real-time requirement was made much more difficult to satisfy by the fact that the algorithm thus had to be executed by the main processor of the system, a Z-80 running on a 4 MHz clock, which also had to take care of all interrupts interrupts from other elements of the system, including data input from the analog-to-digital convertor, and also control the CRT display. As a result, less than 50% of the computing power of the main processor is available for the execution of the pitch extraction algorithm employed. This led the author to develop a new highly computationally efficient time domain algorithm combining a data reduction and parallel processing approach [4]. The only speaker-dependent parameter required by the algorithm is an estimate of the speaker's normal pitch range, but this value need be only very approximate, and could easily be enterred simply by specifying the speaker as a man, woman, boy, or girl. It is automatically set when the system is re-initialized for a new user, a process that takes only a few seconds. A timing analysis performed for the case of voiced speech indicates that the algorithm requires about 0.75 sec of computation time for every second of data analyzed, where the estimated pitch is updated at each new pitch period, and the speech is sampled at a rate of 10 KHz. Thus, for continuous pitch training, a frame oriented approach must be adopted.

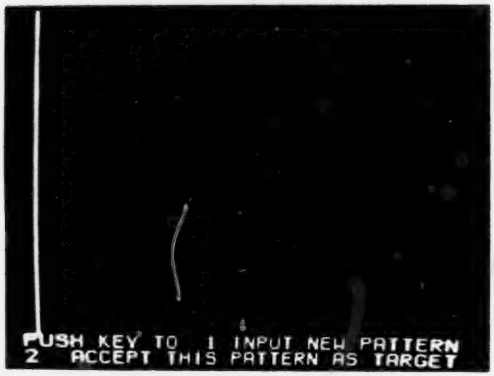
Operating Modes and Training Methods

In normal operation the system functions as a completely independent, stand-alone system. However, it also has the capability to be interfaced with a personal computer to enable the collection of data, and in addition giving it the the potential to utilized in a computer-aided instruction system. It is also possible to use the computer as a source of additional training targets, although the system already has the capacity for a wide variety of targets even in the stand alone mode. The system functions in four different training display modes as described below.

Intensity Training. In this mode, the intensity of the input speech signal is measured as the short-time average energy computed for overlapping 102.4 msec intervals, thus providing a relatively smoothly changing estimate of the intensity of the signal which is updated about 20 times per second. These estimates are displayed as the height of a cursor moving horizontally from left to right across the screen. Past positions of the cursor are not erased, so at the end of a single training, a trace is left on the CRT screen indicating how the intensity varied over the training period. The horizontal speed of the cursor can be adjusted to vary the total duration of the training period as desired. The display can be used in this way simply to provide feedback to the trainee regarding the intensity of his or her speech output, or a target can be displayed which the trainee then attempts to match. This target can be one of five predefined sets of lines determining a series of changing levels within whose outer boundaries the user attempts to stay (See Photo 1). It can also be one of up to six intensity trace targets which can be stored in the system by the therapist before the start of or during a given training session. These give the system great flexibility, and allow the therapist to provide targets at an appropriate level of difficulty for a given student, as well as providing targets which correspond to natural speech patterns and allow the student to practice meaningful intonation patterns. The predefined line patterns enable the student to practice with the system even when no one is available to provide other targets.

Photo 1. Intensity training display.

Pitch Training. The format of this mode is identical to that of the intensity training mode, except that the height of the cursor reflects the estimate of the fundamental frequency of the input speech signal determined by the pitch extraction algorithm. This estimate is updated about fifteen times per second, rapidly enough so that the cursor trace displayed is a relatively smooth representation of the pitch contour. In addition, the intensity of the speech input is simultaneously



PUSH KEY TO 1 INPUT NEW PATTERN
2 ACCEPT THIS PATTERN AS TARGET

Photo 2. Pitch training (contour display)

computed so that a message can be displayed when the intensity rises above a set threshold. When the pitch ranges of the therapist and the trainee differ, the display of the target model is offset appropriately to account for the difference, so that the model can always be generated by the therapist in his or her own naturally pitched voice. (See Photo 2 for an illustration of such a pitch contour display).

Vowel Training. In this mode, the cursor appears on the screen in a position which reflects the values of the first and second formants extracted from the spectrum of the speech signal, where the first formant is displayed horizontally, and the second formant vertically. The position of the cursor is updated sixteen times per second, and disappears during silent intervals to avoid the appearance of distracting, meaningless points in the display. There are two cursor display modes for vowel training: single-point and trace mode. In the single-point mode, the previous cursor position is erased each time a new point is displayed. This enables the trainee to use the feedback from the display to "home in" on the pronounciation of a particular vowel, adjusting his or her speech so that the cursor moves to that area of the F1-F2 plane corresponding to the target vowel chosen. These target areas are marked on the display with the names of the vowels (See Photo 3. At present, the system is being developed for use in Japan, so only the positions of the five vowels of the Japanese language are marked). In the trace mode, the previous positions of the cursor are not erased, so that the resultant display indicates the first and second formant transitions which took place during the utterance. The update rate of the cursor is sufficiently rapid to enable the trainee to practice the pronounciation of dipthongs. In addition, in the case of the Japanese language, the normal rate of speaking is about five syllables (and thus five vowel sounds) per second, so that more advanced students can use this mode to check on the accuracy of their pronounciation of vowels in words and word combinations.

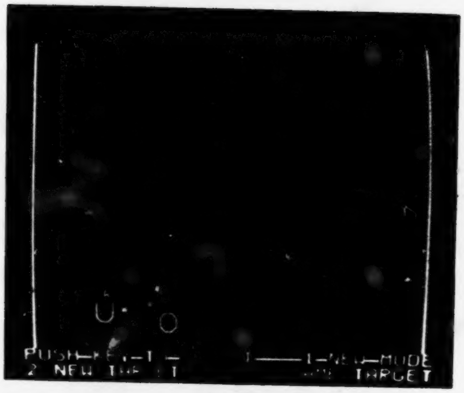


Photo 3. Vowel training, trace mode. for Japanese word "ue" (OO-E).

CONCLUSION

In this paper, the design and method of utilization of a speech training system for the hearing impaired has been presented. The system can provide useful feedback in real-time on the intensity, pitch level, and first two formants of the speech output of the trainee. This system should prove to be a valuable tool, enabling both the speech therapist and the trainee to make more effective use of their time. This system is currently being field-tested, and at the same time a new, more advanced system is being designed and built with the goal of of providing training in the pronounciation of consonants as well.

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19.3

Development of a Portable Voice Output Communication System with Prosodic Feature Control

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ABSTRACT

A portable communication system was designed and implemented which incorporates text-to-speech capability, user controllable prosodic features, compact size, portability, text storage, and accelerated text entry. The resulting system allows sophisticated control of the synthetic voice through single finger typing.

INTRODUCTION

We have developed a portable communication system which is optimized for conversational use. The system consists of the component parts shown in the figure. Emphasis has been placed on maximizing communication rate via imput acceleration and maximizing speech output quality through user control of prosodic features.

BACKGROUND

Human speech makes use of changing pitch and amplitude, variable syllable length, and changing voice quality to lend meaning and interest, variations which we refer to as "inflection", or prosodic features. This system uses the SSI-263 is a phoneme based speech synthesis chip. The SSI-263 can produce as output (that is, speak,) a finite number or phonemes, which can be linked together to form words. A letter-to-phoneme algorithim performs the conversion from input text to phonemes. The simple act of stringing together phonemes, however, does not produce high quality speech. The SSI-263 has eight parameters that can be changed to simulate the prosodic features of human speech. These are: inflection level, rate of inflection change, rate of articulation, rate of speech, amplitude, filter frequency, duration of phoneme and phoneme. The use of these eight parameters can greatly improve the quality of the synthetic speech the SSI-263 can provide. To control all of them individually, however, is not really practical. Our goal has been to allow the user the greatest possible flexibility with the least possible effort.

The control of prosodic features necessary in a prosthetic voice differ greatly from the controls needed in other devices which use synthetic speech. In a device with a limited, situation-specific vocabulary, for example, all inflection can be stored as a part of the utterance, and therefore, no special control characters are needed. In the prosthetic voice, however, it is important that the user be able to control the inflection. The controls must be inserted into the text by the user,and must, therefore, have easily understood functions that are meaningful to the user.

PROSODIC FEATURES PROGRAM

Our first step in determining how these parameters can best be used to improve voice quality was to write a program which enabled us to set each of the eight parameters individually for each phoneme of an utterance. In this way, we determined which parameters were most useful, and which could be left at a single, default, value, as well as what that default value should be. We also found how best to use the parameters we did want to vary.

We began with a number of test sentences, which we entered into the computer phoneme by phoneme. Then we experimented with the parameters until the utterance sounded as natural as we could make it. We used such sentences as, "Are you leaving, now?" and "They're married." In the first sentence, we tried to make the utterance sound like a neutral question, and then we emphasized different words to change the meaning slightly. In the second sentence, we tried to make the utterange sound like a statement and then like a question, "They're married?"

Next, we looked at words such as "content," which have a different meaning, depending upon which syllable has been stressed. We tried to make "content" sound like a noun and like an adjective.

Finally, we analyzed what we had done in terms of fairly generalized rules which we could then apply to other sentences and words. The next step was to devise a way to apply the rules; that is, a way to control the prosodic features.

We decided to use two approaches to prosody control. One approach was to use what we termed "automatic inflection." The other was to make limited use of hand-placed stress.

Automatic inflection means that inflection is added to the sentences on the basis of the punctuation used. We started out with two sentence patterns: a statement pattern with falling pitch at the end, and a question pattern with rising pitch. The pitch of the sentence varies from the default for the first few phonemes and the last few phonemes of the sentence. After some experimentation with the first two sentence patterns, we decided to add two more statement patterns, as a single pattern soon becomes boring. Thus, we have the three statement patterns and one question pattern, which are imposed on the utterance by the user's insertion into the text of one of the following punctuation marks: (.), (:), (;), or (?). The rules governing these sentence patterns operate on the pitch values of the phoneme codes resulting from the translation of the current section of text. Sections of text are delineated by these punctuation characters: (.), (:), (;), (?), (!), and by carriage returns. Because the patterns affect only the first and last few phonemes of a section of text, they are most effective with short sentences, or when the user breaks up the sentences into shorter phrases.

In the hand placed stress method, the user can choose to emphasize a particular word within a sentence, or a particular syllable within a word. The user emphasizes a word by placing an asteric (*) at the end of the word, either in place of, or before, the space; a syllable is emphasized by placing a plus sign (+) after it in the word.

As an example, the user might type, "The plain fact of the matter is, they have not succeeded." The pitch would change some at the very start of the sentence, and again at the end, during the last syllables of "succeeded," but the entire middle of the sentence would be quite flat. Suppose, however, that the user typed, "The plain*fact*of the matter is, they have not*succeed+ed." The text of the second sentence contains only one more character than that of the first, but the utterance produced by the second sentence contains a great deal more character.

TEXT ENTRY HARDWARE

The system is designed for those who can directly select keys on a typewriter keyboard, and have a broad range of communication goals in terms of spoken and printed output. There are many computers currently available which have the features necessary to implement a text entry and manipulation program. The eventual choice of a computer that would allow effective text entry was based on several criteria. The computer chosen must allow the inclusion of a text entry acceleration program that interacts with the built-in features of the computer in an effective manner. The system must have I/O capabilities which allow it to interact with serial and parallel peripheral devices such as printers, other computers and the voice synthesis unit. The computer chosen should have an adequate display to permit useful text editing. Finally, the system must be lightweight, portable, and have a power supply sufficient to permit all-day use without recharging.

A review of all the currently available portable notebook computers (1) resulted in the choice of the NEC 8201A computer. This notebook computer displays 40 (0.25 inch high) characters in 8 lines on an LCD display, the minimum size necessary for effective text editing. Although units with larger LCD displays are available, they present characters which may be too small for many users to see adequately, and are vulnerable to damage due to their "swing open" construction. This "swing open" construction also makes the display of these units difficult to open for viewing for many potential users. The standard keyboard of the NEC features 10 (5 plus 5 w/shift) programmable special function keys and particularly well-laid out editing keys. This latter feature is extremely important to users who are apt to need frequent editing to produce mistake free text. The NEC is configured with 32K bytes of ROM and up to 96K bytes of RAM which can be addressed one 32K bank at a time. Data may be transferred from bank to bank with a machine language utility. An internal battery will maintain the information stored in RAM for up to one month without recharging, even if the display batteries are fully discharged. Bundled software contained in the ROM of the NEC includes a powerful text editor, a complete version of BASIC and a powerful operating system that handles file manipulation and I/O processes. Memory may be configured so that machine language programs can be created and stored in files which offer minimal exposure to accidental modification or erasure. The overall programming environment allows the effective implementation of a text entry acceleration program.

TEXT ENTRY PROGRAM

A program which implemented an abbreviation/expansion routine (2) was developed to reduce the number of keystrokes necessary to produce text. The program was written in BASIC, with machine language subroutines used to implement time critical functions such as display updating, scrolling, text buffer erasure and various search functions. To save memory, machine language subroutine calls were made to pre-existing routines stored in the NEC standard ROMs when possible. This approach saved a significant amount of memory and was very useful for implementing the I/O routines utilized by the program. A menu-driven 10-Branch abbreviation/expansion software package was developed which allows text entry, and scans the text as it is entered for character strings with single number suffixes which are used to to identify abbreviations. This program functions essentially as a text editor that examines all user input for abbreviations. If an abbreviation occurs, it is quickly and automatically expanded into a complete word or phrase. Text may be entered, edited and stored in RAM memory, and later be dumped to the speech synthesizer, an external storage device or printer.

The program also includes a number of special characteristics designed to enhance the ease and speed of text entry. The text produced by the abbreviation/expansion program may be stored in a form that allows interaction with the built-in text editor. This allows the powerful resident word processing capabilities of the NEC to be utilized to cut and paste, delete text, and so forth. Pressing the ESCAPE key accesses a menu which lists all the special features to edit and list abbreviations and expansions, change program parameters and perform various file manipulations. Since the SHIFT key cannot be separately detected with software, the TAB key has been modified by the program to produce uppercase characters. If the tab is struck once, the next key depressed will result in an uppercase character. If it is struck twice, all following keystrokes will produce uppercase characters, until the tab key is pressed again, which will cause the output to revert to lowercase. The special function keys on the NEC 8201A have been redefined to perform such functions as blanking the text buffer and entering new text, allowing the cursor to be moved anywhere in the text, speaking the contents of the buffer via the speech synthesizer, previewing spoken text, etc. Many keys, such as the "*", "/", "\", "[" and "]" keys are used to cause special features of the speech synthesis software to occur. These keys are all available without shifting. The square bracket keys, [and] can be used to selectively send text to the speech synthesizer. This feature allows conversation during editing of a separate file. Only bracketed text will be output to the speech synthesizer.

Abbreviations and expansions utilized by the system's operator can be listed, added, changed or deleted at the option of the user. The program is always resident in the RAM memory of the computer, and has been configured to automatically run when the NEC is turned on. The development of this system illustrates the use of standard hardware with a custom program to increase the rate of text entry (abbreviation/expansion).

CONCLUSIONS

Portable microcomputers can serve important functions for disabled individuals if adequate assessment of needs and skills is carried out prior to deciding on a system, adequate training and documentation are included and necessary followup services are available.

The system described here addresses the unique requirements presented by conversational needs of non-speaking clients.

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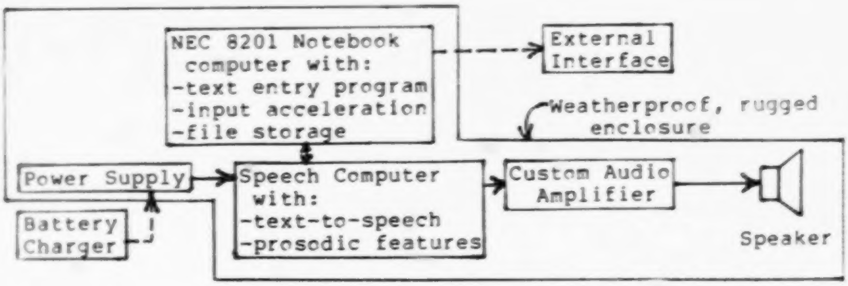

Block Diagram of Custom Portable Speech Output Communication system

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EYE FOR INDEPENDENCE

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ABSTRACT

To a disabled person, independence means being an active participant in life, able to perform a wide variety of tasks and functions without another person's help. To achieve this, he or she requires an ability to communicate and control. A system for non-invasively measuring a person's line of sight has been combined with a communicator and controller, to enable severely disabled persons to communicate at speeds up to 45 words per minute in both conversational and written modes and to independently control or operate a wide range of assistive devices as well as standard computer based facilities. With it, their independence, and their educational, vocational, recreational and social opportunities will be significantly increased.

INTRODUCTION

A person's self worth, and perceived usefulness to society is intimately linked to the level of independence which he or she enjoys. To become and remain independent, a person needs to reach high levels of independent communication and control, as it is these basic skills which will determine the level of a person's participation in all activities and the extent and the speed of his or her performance in attempting complex tasks and functions. Once acquired, communication skills allow most people to read silently at speeds up to 250 words per minute (w.p.m.) and to converse at speeds in excess of 150 w.p.m. Written expression, in composition mode, is considerably slower at about 20 w.p.m., but in most circumstances this is still acceptable, as there are generally other long subsequent delays before the message is read. Impatient computer terminals, however, are now making us far more aware of our short-comings in this area and faster alternate input methods are continually being sought. Perhaps we can start to imagine the frustration at being able to express oneself at no more than a maximum of 5 w.p.m. in any mode.

METHODS

Speech is the fastest, most preferred form of communication, but while many non-vocal, physically handicapped persons including high-level quadraplegics, and persons with cerebral palsy, amyotrophic lateral sclerosis or other neurological disorders have normal intelligence, can listen and read normally, and formulate responses, they lack the ability to express themselves. A wide range of communication and control aids have been devised for such persons, but despite best efforts, the resultant overall performance often remains unsatisfactory. An interpreter familiar with the user's techniques is still required, and expressive communication is stilted and very slow. The topics and the greater proportions of a conversation are dominated by others, with listeners anticipating responses and asking questions requiring short answers in order to speed things up. Turn taking rules are ignored and the speech impaired person has limited opportunities to initiate conversation. For disabled students, practice, revision and question times, essential in a normal educational process, are often inadequate for satisfactory progress to be made. Educational, vocational, recreational and social opportunities are all greatly restricted. The full potential of speech impaired persons is far from realised, and they in fact become increasingly dependent on others for many of their needs. A reversal of this situation can only occur if basic communication and control skills are significantly increased.

An eye-gaze communicator and controller

A system has been devised whereby speech impaired persons with poor motor control can operate a communicator, environmental controller or other computer based systems by eye fixations. The complete system comprises a microcomputer controlled eye monitoring system for non-invasively determining the user's line of sight, and a screen-based communicator and controller. Interfaces enable the system to drive peripheral devices such as a printer or speech synthesiser, either directly or via another computer. Eye-gaze control was considered for two main reasons. Firstly, many severely disabled persons retain good control of their eye movements, despite other motor skills being severely impaired, and can fixate quite readily upon a target such as a word or control function in a display matrix. Secondly, in most types of communication and control aids employing direct selection techniques, eye contact with the intended selection occurs some time before it is implemented by some other physical action of the user. If eye control is used, selection can occur as soon as the system is satisfied that a particular word or character has been fixated upon for a sufficient time of the order of half a second. Eye control is potentially the fastest selection method available for persons requiring a communication aid.

The eye-gaze monitor. This acts as the principal input transducer for the communicator. Numerous methods have been devised to monitor eye movements (1), and several forms of communicator based on them are being developed elsewhere. Rinard et al (2) use the corneal reflection method to measure eye position relative to the head and have automated the eye-gaze interpretation process in the ETRAN communicator. Free head movement is allowed, and no fixation target is required once the user has been trained, but the system is intrusive, with special apparatus attached to spectacle frames worn by the user. A maximum speed of 12 w.p.m. could be expected. Foulds et al (3) measure eye-head position in a similar way, but, in addition, use an ultrasonic ranging technique to determine head position and orientation. The two sets of measurements together enable absolute eye gaze to be determined to about 1 degree. Free head movement is permitted, but again parts of the system are attached to the user, and may necessitate frequent, lengthy setting up and calibration procedures, so that the continuous or frequent presence of an attendant is required.

Our system is similar to that described by Merchant et al (4), but without the eye tracking facility. The user is seated comfortably about 60 cm from the communicator's TV display, with gentle lateral head support to allow limited movement while remaining within the field of view of a sensitive TV camera focussed on one eye. An important feature of this system is that the user is completely detached from the monitoring system. A low-level light source is directed at the user's eye along the axis of the camera, illuminating the retina so that the pupil is seen back-lit as a bright ellipse. The corneal reflection of the light source appears as a small, very bright spot near the centre of the eye. If the optical axis of the user's eye were directed at the camera, the reflection spot would be at the centre of the pupil. As the gaze is shifted laterally or vertically, with the head still, the corneal reflection shifts in the same direction as the pupil, but to a lesser extent due to the shape of the cornea. The length and direction of the vector distance from the centre of the corneal reflection to the centre of the pupil is a measure of the user's line of sight and is largely independent of head position. Signal processing hardware and software within the system accurately determines the positions of both features, and enables the point on the target area at which the user is looking to be predicted at a rate of 30 times per second. Prior to use, the system must be set up and calibrated, with an attendant required solely for aligning the user with the apparatus. Thereafter, all functions are controlled by the user, and calibration is done by the user looking at specific target points on the display for about 1 second each while the eye gaze monitor calculates an average value of the pupil-reflection vector for each point. The setting up procedure takes only a few minutes. Because none of the apparatus is attached to the user, the likelihood of it being physically disturbed is very small and recalibration is seldom necessary.

After calibration, and when the user is looking at some arbitrary point on the display screen, an interpolation algorithm is used to calculate the position of this point from the observed pupil reflection vector. The system is able to resolve the direction of eye-gaze in both azimuth and elevation to within one degree. At a viewing distance of 55-60 cm from a 30 cm visual display unit, the screen subtends angles of 24° horizontally and 18° vertically, and each rectangular character, control function or word cell in the communicator's display matrix will subtend angles of 2.4° and 1.5°. As a consequence, the eye-gaze monitor can unambiguously determine which cell is being fixated by the user.

The communicator. This is operated by the user interacting with the dynamically alterable TV display screen via the eye gaze or other appropriate cursor control transducer such as a joystick, light-pen or mouse. The screen uses a standard format of 24 rows of 80 characters and a full set of ASCII characters is available. Typically, the top 6 rows form a message assembly area for the user, and the bottom row is used for user operating instructions or diagnostic information. Rows 7 to 23 display one of the 25 pages of a dictionary of characters, words, phrases and control functions. This large store, with a present capacity of about 1600 entries, enables the communicator to be word rather than character oriented, with a corresponding increase in its potential speed for any of the transducers used. The page approach overcomes difficulties in attempting to make direct selections from two dimensional displays containing a large number of words where a high degree of pointing ability is demanded of the user.

The active area of each page is arranged as a regular matrix of 8 rows of 9 cells, with each cell having 8 character spaces including a leading blank. 94% of the words required in the dictionary had 7 or less characters. Longer words, and phrases up to 70 characters in length may extend over several cells. Alternatively, an abbreviation-expansion technique can be used to maintain a large number of entries per page. Most of the dictionary was arranged alphabetically by columns and left justified to facilitate word searching by the user. The separation of the word columns is less than is normally recommended for screen-based lists (5), but is being tried in order to exploit the attainable accuracy of the line of sight measurements and to maintain a high communication rate by making more entries directly accessible.

The selection of pages, or their contents, is achieved by the user looking at the desired cell in order to position an inverse video cursor over it, and then either continuing to fixate it for a short, user-adjustable dwell time, or operating an auxiliary switch, in order to confirm his selection. The whole of the selected cell changes to inverse video momentarily to provide visual feedback to the user, acknowledging his selection, and the communicator executes the required action such as writing a selected character or word into the message assembly area, or displaying the new dictionary page. No selections can be made if the cursor is located in the "rest" areas surrounding the active region of the screen.

The index page is the most important page in the dictionary, and the communicator generally returns to it automatically after each word selection process is complete and an inter-word space inserted. 16 of its cells allow the user to quickly go to another page in the dictionary with a first selection step, and then with a second step to choose a word from that page to complete the selection process. The 48 remaining cells on the index page are occupied by frequently used words which will only require one step to select them.

An additional column of 8 cells is common to all pages, and contains important control functions which must be accessible from every page. The first of these is an ON/OFF toggle switch which may be used to temporarily disable every selection cell, except itself. This facility is essential if an automatic selection procedure based on dwell time is used, especially if the communicator is controlled by eye gaze. With the communicator disabled, the user can safely inspect the contents of the current page or practise moving the cursor, without the fear of making an inadvertent selection. Other cells in the control block provide immediate access back to the main index page, and to pages of alphanumeric and other single characters including punctuation, common affixes, and phrases. A larger cell in the control block enables the user to completely cancel the effect of a last incorrect selection.

The vocabulary. An oral vocabulary, based on a 1956 survey by Schonell et al (6), was chosen deliberately as the communicator would be used principally in a conversational mode. The first thousand head-words, in order of frequency of occurrence in a total of 512,647 spoken words, were further broken down into the distinct word forms necessary if they were to be included in our dictionary. For example, the head-word *has*, includes the words forms *has*, *have*, *having* and *had*. It also includes *'s*, *'d*, and *'ve* as used in elisions such as *he'd* and *I've*. The expanded list was re-sorted into decreasing frequency order, and truncated at 825 words. The relative occurrences of all words spoken in the survey decreased rapidly from a maximum of 3.82% for the most common word, *I*, to 0.40% for the 48th word *are* and 0.007% for the 825th.

The first 48 words, incorporated in the communicator's index page, collectively account for 51% of all spoken words, and the remaining words, stored in 16 pages of the body of the dictionary, account for a further 37%. The remaining 12% of the words to be selected would need to be spelt completely, requiring (n+2) steps for an n-letter word, or constructed from other part or whole words in the dictionary.

Further enhancements have been made so that the system is also well suited to written forms of communication, and pages of control functions have been added to enable peripheral devices such as environmental controllers, computers, word processors etc. to be easily operated. Early users of the system found that the most common words of the average adult did not necessarily suit them, and several insertions and deletions of words were made in each case. It is clear that some personalisation of the dictionary is required, and it is estimated, for a dictionary of about 1000 words, that 90% of words used would be directly accessible and that an average of only 2.1 selection steps per word would be required. The most appropriate size of dictionary will depend on the mental skills and language proficiency of the user, but the low usage of additional words makes it unlikely that an increase in dictionary size much beyond 1000 words will provide a significant increase in communication speed or efficiency. Users would also have increasing difficulty remembering the position on a page of the less frequent words in an extended dictionary, increasing search times and hence selection times. With the present size of 1000 words, and with the user familiar with the word placements, it is anticipated that it would take approximately 0.6 seconds to perform each selection step. Assuming 2.1 selection steps per word, this is equivalent to a communication rate of 45-50 words per minute, and represents a 10-20 fold increase in communication speed for severely disabled users.

CONCLUSIONS

The eye gaze communicator and controller will be able to significantly enhance the capabilities of severely disabled persons. With faster communication rates, students will be able to reach higher levels of education in a shorter time, and accident victims can be rehabilitated more quickly. The system can provide ready access to word processors and, in fact, any standard computer hardware and software, opening up a wide range of vocational opportunities, with the possibility that the user could become gainfully employed and a taxpayer. Many daily living activities could be undertaken with its assistance as a controller of other devices, including environmental controllers and robotic aids. Lesser amounts of attendant care would be required. All of these aspects lead to a higher level of independence and greater self esteem for the disabled person, and to a significantly reduced financial cost to the rest of society.

ACKNOWLEDGEMENT

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19.5

TOWARDS COMPUTER-AIDED VISUAL COMMUNICATION *FOR APHASICS: REPORT OF STUDIES*

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ABSTRACT

Since July, 1984, researchers at the Palo Alto VA Rehabilitation R&D Center and at Stanford University, in consultation with Boston colleagues, have been conducting pilot studies to investigate the feasibility and utility of implementing a visual communication system for aphasics on a computer with graphic display. These studies have replicated major findings of work done earlier in Boston, while suggesting ways of adapting the symbolic communication system to the computer environment to improve ease of use, increase vocabulary size, and augment expressive power.

BACKGROUND

In the mid-70's, teams of neurologists and aphasiologists in Boston[1,2] and New York[3] investigated using visually based communication systems for aphasia remediation. The studies were inspired by earlier successes in training infra-human primates to communicate using material tokens and sign language. The work of the two groups used broadly similar systems, based on manipulable tokens, but employed different training regimens: the New York group adhered to strict operant conditioning, while the Boston group — calling its system VIC for VIsual Communication — used a naturalistic approach. The populations trained at the two centers comprised groups of similar size (7-8), all of whose member had sustained left hemisphere damage. Most of the subjects were globally aphasic.

Both research groups found subjects who could learn to use the visual communication system and employ it in selected communicative tasks. The Boston group states:

> The evidence suggests that some severely aphasic patients can master the basics of an alternative symbol system. Moreover several indices suggest that the communicative consequences of the system are appreciated, and that at least some of the cognitive operations entailed in natural language persist despite severe aphasia. (Gardner, 1976, p. 275)

Those investigations also showed that: 1) pre-screening of subjects for specific relevant competencies is required; 2) more than half the screened subjects (five of eight) could master the basics of VIC; 3) error rates among successful trainees were low; 4) patterns of error were similar across subjects, and stable; 5) two of the subjects learned to express some of their feelings and desires using the system; and 6) performance in VIC far surpassed performance in English on equivalent tasks. Fig. 1 shows a simple VIC communication, which means "Touch the spoon, JN!".

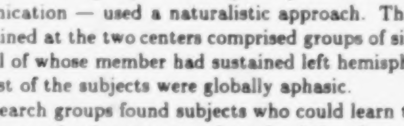

FIG. 1

Despite positive clinical findings, VIC was almost never used by subjects outside training sessions. Implemented as icons drawn on index cards, VIC was found to be cumbersome, demanding to use, and of limited practical utility. Subjects required a clear table top to arrange card piles and communications; accessing cards from stacks was difficult for right-hemiparetic subjects; and vocabulary had to be limited to keep card stacks at manageable size.

The current investigations in Palo Alto continue the Boston line of work, both in the naturalistic approach to training, and in many specific features of the communication system proper (e.g., icons, lexicon, syntax). What modifications have been introduced are aimed at adapting the system to the computer environment for aphasic users; subsequent modifications can be expected as the principles of visual communication in a computer-graphic environment are refined. The immediate goals of this work are: 1) to explore, extend and refine the principles of visual communication in aphasia; 2) to establish to what degree, and in what ways, a computer with high-quality graphic output can compensate for the difficulties encountered earlier; and 3) to replicate the major findings of the earlier studies in the non-computer medium. In the longer term, we would like to contribute to the development of a practical, portable device, incorporating a graphic interface for visual communication, which aphasic persons can use in simple functional communicative tasks.

CURRENT TRAINING PROGRAM

Two subjects are currently involved in our pilot study. While both manifest severe expressive aphasia, with somewhat less severe comprehension deficits, their histories are significantly different. One subject, JN, is a 44 year old, right-handed man, who was in good health until 1974, when he suffered a head injury in a motorcycle accident and subsequently underwent a partial frontal lobotomy. Since that time, JN has suffered from a severe expressive aphasia. Despite great enthusiasm on the part of the patient, all attempts for speech therapy have been unsuccessful. The Boston Diagnostic Aphasia Examination revealed subscores near or at the zero level for naming, reading and repetition. Inspection of the Profile of Speech Characteristics shows that fluency in free speech, including articulation, is profoundly impaired. Melodic line is absent. Scores for phrase length, grammatical form, paraphasias in running speech and word finding could not be obtained due to impaired articulatory agility. Auditory comprehension is relatively less impaired, with a Z score of -1. Spontaneous written expression is confined to highly routinized sequences, such as the patient's name and address, and the patient exhibits moderate ability to copy, occasionally replacing printed forms with cursive script forms. There is no comprehension of written material.

Our other subject, RB, is a 55 year old right handed man who sustained a middle cerebral artery infarct in September, 1982, resulting in a severe right hemiplegia, a mild apraxia, and global aphasia. Early spontaneous recovery, coupled with speech therapy, partially restored auditory comprehension, and RB now responds to yes/no questions with between 40% and 75% accuracy. His expressive deficits, however, remain severe, with only a few stereotyped utterances produced spontaneously. Examination reveals no useable word finding ability, no running speech, no facility for grammatical form, phrase length, or melodic line. Performance in naming and reading is profoundly impaired. He can repeat words and short phrases with inconsistent success, though he appears not to comprehend the meaning of what he is saying. He displays a moderate ability to copy directly, with considerable effort, but exhibits no comprehension of written material.

RESULTS OF WORK WITH CARDS

These subjects, despite comparable performance on screening examinations, have demonstrated distinctly different abilities in acquiring the VIC system. JN has progressed as rapidly as the average successful learner in the Boston VIC work. Moreover, his performance is similar to the Boston subjects' in other respects: comprehension and expression in VIC are consistently better than in English; he has mastered three major communicative functions which the VIC system was specified to subserve (e.g., following commands, describing ongoing events, answering questions); his performance strengths and weaknesses parallel those of the Boston subjects; he appears to perform best after "warming up" in the session. RB, on the other hand, has shown difficulty in acquiring the concept of a visual syntax: he can identify individual cards accurately, but inconsistenly associates strings such as *transitive verb plus noun* (e.g., [SHAKE][BOX]) with the appropriate actions. Such problems have been noted by Luria, who categorizes them as a *simultaneous synthesis* difficulty. They arise from a deficit in grasping the logical-grammatical relations between word combinations, and lead to an inability to comprehend and produce word groups requiring complex simultaneous organization.[4] Experimental approaches are being explored to assist RB, but the outcome is uncertain. The situation suggests that additional screening tests, yet to be identified, may prove useful in VIC candidate selection.

Analysis of JN's successful performance data reveals some interesting facts. First, frequency of accurate icon selection is a function of grammatical category: common nouns and proper nouns are selected accurately most frequently, verbs come next, and prepositions follow. Second, ease of item acquisition from different grammatical categories shows a similar profile: nouns are assimilated practically immediately, verbs must be rehearsed several times, prepositions in practice must be refreshed at the outset of each training session. Third, the data show that both intensive drills and warming up periods are integral parts of the training: the intensive drills establish the basic patterns, warm-ups refresh the subject's memory prior to use during VIC sessions. Fourth, the work highlights the importance of identifying general principles of iconic visual communication and incorporating them into VIC. English syntax, the original model for the VIC grammar, is derived from communication via a non-visual channel, and thus provides only partial guidance in this task. Fifth, JN's performance on a preliminary laboratory computer-VIC suggests that operational demands do not exceed his abilities. In his first exposure to the system, he quickly mastered key cognitive and motor tasks associated with operating the interface, and quickly set about the exploring the capabilities of the system.

TOWARDS COMPUTERIZATION

Computerization of a visual communication system is an ambitious project, and one which will require time and effort in several substages. Our current attentions are focused on two sub-projects, which allow us to utilize currently available resources.

A Research Implementation

One of the authors, F. Lakin, working on a Symbolics 3600 LISP Machine at SRI International, has implemented a general-purpose Visual Communication Laboratory.[5] This Laboratory makes it possible to implement and modify versions of VIC quickly and easily on the computer. In addition, it automatically records a history of user interaction, making later playback and review available for analytical and diagnostic purposes.[6]

On August 1, 1984, JN was given a first training session on the computer implementation of VIC. In the course of approximately two hours, JN demonstrated an ability to master the use of the mouse pointing-device interface to track objects, point to them, and draw new objects. He demonstrated that he recognized VIC symbols on the graphic display screen, that he understood computer-VIC commands, and that he could follow them appropriately. Additionally, on his own initiative, JN constructed his first full VIC communication in any medium on the computer system, demonstrating that he understood the message by carrying out the appropriate action. Later experiments, which tested JN's response to high quality synthesized speech (from a DECTalk unit), showed that his performance in response to synthesized voice output was comparable to that in response to the spoken human voice. Such work is significant in two ways: 1) it suggests that use of high technology aids by aphasics in assistive communicative devices is not in principle precluded, and 2) it gives the researchers valuable experience in designing and testing interfaces for this rehabilitation application.

Explorable Bitworld

A joint VA-Stanford proposal has been submitted to Stanford University for support to develop software of both pedagogical application for language students and rehabilitative application for aphasic persons.[7] The approach is to gain maximum leverage from recent efforts to develop interfaces which are largely intuitively clear to unsophisticated users. Much effort for the business market, for example, has gone into the elaboration of the "desktop metaphor", wherein office workers employ icons (i.e., small, stylized renditions) of familiar objects (e.g., notepads, file folders) to carry out common business tasks (e.g., write memos, process documents). To construct a "Bitworld" is to extend and modify this metaphor approach, in hierarchical modules, so that the user can eventually move about the room as well as the desktop, around the home as well as the room, and around the neighborhood as well as the home. The Macintosh computer, because of its compact size, relatively low cost, mouse interface (useful in circumventing the frequent hemiplegia of aphasic subjects), high-resolution display, and excellent graphics ROM routines, is currently considered the most attractive target machine.[8]

The specific objectives of the project are: 1) to design software for the Macintosh computer which allows the display of a Bitworld, through which the user can navigate as he/she chooses; 2) to organize this Bitworld to be conventional in layout and appointments, filled with common, recognizable and useful objects; 3) to provide the use with the capability of "selecting" some object of attention; 4) to provide software which will then respond in some specified fashion (e.g., by writing/speaking the name of the object); 5) to implement these capabilities in as natural a fashion as possible, using the hardware (e.g., mouse) and software techniques (e.g., pull-down menus, "opening" icons) developed for the Macintosh; 6) to test the utility of this software and hardware in producing a VIC which is more powerful, natural and expressive than the card-based VIC; and 7) to revise the software, improving the utility of the system on the basis of tests by the target population. If Stanford approves and supports this project, we anticipate having a preliminary version of the Bitworld available for testing by the beginning of the upcoming academic year, in September 1985.

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19.6

INTERACTIVE EVALUATION OF CURRENT VOICE INPUT TECHNOLOGY FOR THE PHYSICALLY DISABLED SUGGESTS NEW AVENUES FOR CONTINUED RESEARCH

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ABSTRACT

The advent of human voice recognition devices at reasonable cost has provided a revolutionary control and communication modality for severely physically disabled individuals. In practice, however, such systems have met with only limited success and varying acceptance. Rigorous objective examination of the human-machine interface in the voice-controlled VA/Stanford University Robotic Aid using an Interactive Evaluation model has highlighted major limitations of present voice-input technology and its practical implementation. These limitations and an approach to resolving them are the subject of this paper.

INTRODUCTION

The present unprecedented increase in numbers of older and frailer persons in our society will continue to change this country's demographics and impact on future health care delivery. Independence and maintenance of vitality and quality of life will require alternative and updated methods for providing suitable services and devices to those with physical limitations. Technological assistive devices have the potential for augmenting human capabilities and providing increased independence, not only to the disabled but also those who care for them (3). Many mundane activities of daily living may be adequately performed by high-tech assistive devices–reserving for nursing staff and family members those functions uniquely provided by humans. The potential saving of time and costs, as well as the psychological benefits accruing from independent living cannot be overemphasized.

BACKGROUND

The converging technologies of machine voice recognition, microprocessors, and robotics have provided a new and exciting stimulus for the development of devices to replace lost or limited function.

The Palo Alto Veterans Administration/Stanford University (VA/SU) Robotic Aid Project has, for the last five years sought to provide proof-of-concept and *feasibility* for the use of industrial hardware in a human service application. This has been successfully implemented and reported on (6). However, only recently with the application of the Interactive Evaluation (I-E) Model has the clinical utility of such a device been both qualitatively and quantitatively investigated and documented (4,5).

This I-E model has provided valuable baseline information about the needs of a wide range of individuals of widely varying abilities and ages (5 to 90). To date, more than 100 subjects have been exposed to a structured "hands-on" learning process designed to assess the acceptability and utility of a voice-controlled robotic arm configured to augment their performance of activities of daily living (1).

This paper does not address the effector (i.e., robotic arm) aspect of the problem which is being exhaustively attacked at Industrial, Aerospace/Military, and Rehabilitation levels (8,11,12). The clinical evaluation of the VA/SU Robotic Aid Project has served to define the advantages as well as the limitations of first-generation speech recognition and synthesis equipment at *utility* level. We will not, in this paper, further discuss issues in speech-synthesis technology except to note that many users have expressed the wish to be able to choose the gender of the "voice" with which the system addresses them. The latest speech synthesizers do have this capability and can therefore adequately address this need in the future (10).

Before examining the *limitations* of speech recognition technology, it may be important to emphasize the conceptual and functional *strengths* of this control modality. Even for those of normal ability a conventional alphanumeric keyboard presents an unfamiliar face for communication tasks. For the elderly and those of diminished learning, visual, or motor ability, this barrier can be perceived as formidable. Voice input as an alternative control modality offers several advantages:

1. It is a natural means of communication.
2. In certain cases of disability, it is the only channel available.
3. It requires no physical linkage (in contrast to a joystick or electromyographic skin electrodes, for example).
4. It offers a wide range of control freedom.
5. By software modifications voice recognition can readily be optimized for the user's personal needs (13).

RESULTS

Qualitative and quantitative evaluation of voice recognition technology using I-E methodology has served to highlight many previously unidentified problems and limitations at various levels:

The inability of the current voice recognition devices to recognize the same words repeated with different pitch, inflection, speed, loudness, phonation, and connectedness, imposes excessive demands for consistency in human performance for general acceptance (5). Voice Recognition Units (VRU's) demands on the user are rigorous and unforgiving for the following reasons: VRU's are generally speaker-specific, i.e., they have to be "taught" the unique characteristics of each vocabulary word as spoken by a particular user. This constrains the user to produce utterances which are very similar to those stored during the teaching sequence. Increasing vocabulary size proportionately decreases the chance of accurate recognition (7). Only a limited number of words can therefore be compared at a time. Other possible words are kept masked and only allowed into the active vocabulary by explicitly unmasking them on command. To maintain an adequate recognition rate, a maximum of 20 words are available at one time in the VA/SU system. This constrains the choice of input words to a small subset of the spoken vocabulary. It is more difficult for novice users to maintain consistency of individual word-utterances within and between sessions. This can lead to four categories of misrecognitions, each with different consequences: Rejections, substitutions, out-of-context errors, and invalid command errors. The implications of these are discussed by Engelhardt et al. (5). This problem could be solved by the continuous analysis of performance and the ability to update stored word-templates in an ongoing fashion so that the latest version of the word-utterance is incorporated into its template. This is no trivial task; syntactic and contextual clues will be needed to make decisions concerning template updates (2,9,10).

A second problem is the inability of present day VRU's to take advantage of the syntactic information implicit in formed speech. Single-word staccato commands, processed individually, cannot transmit this useful syntactic information. When humans converse, the ability of the listener to anticipate the next word on the basis of learned grammatical rules has a profound effect on comprehension, especially when individual words are missed due to high background noise levels.This is the so-called *cocktail party phenomenon* (9). By implication, humans in this situation are subconsciously performing a probablistic analysis of the likelihood of a certain word following and/or preceding another in a given sentence. For example, consider this simple syntactic model for a voice-command sentence: subject/verb/object. "You fetch milk" is a typical sequence which fits this model. Even if the speech recognizer heard "chew hedge talc" because of poor enunciation or noise, simple programmed rules of permissible syntax would cause the system to reject this nonsensical sequence and to search its voice templates for a more acceptable syntactic match. This added capability is particularly important when free voice input is used instead of a keyboard for text-typing tasks, since the number of potential misrecognitions in a large vocabulary would otherwise be unacceptable.

Present models of voice/language recognition are too simplistic, since they overlook an even more fundamental property of human conversation,i.e. the semantic and contextual framework of the interchange.When people speak, they do so in a specific context against a mutually understood background.If one adds additional cognitive clues such as place/time,smell and sight, the model of human speech is extremely complex. How to use all available information to make appropriate decisions by machine is the domain of computer science and is the basis of knowledge-based/expert systems (8,10). Due to the unwieldy size of such programs and the limitations of current hardware, it may be some time before these advances are more generally applicable. Nevertheless, we hypothesize that by using even a limited number of these constructs and currently available voice-input hardware, an appreciable improvement in speed, accuracy and vocabulary size in voice recognition is achievable (2).

FUTURE DIRECTIONS

Findings provided by Interactive Evaluation have highlighted at least three limitations of the human-machine voice interface of the VA/SU Robotic Aid project as presently configured (5). Future research on the voice recognition system will be directed specifically to address these deficiencies:

1. Allowing short strings (2 or 3) of permissible command words will permit syntactic analysis. A high degree of determinism will thus be implicit in the command sequences, which should lead to an improved recognition rate over words examined singly.
2. Online performance analysis will enable continuous updating of word-templates. This will ensure that these templates are the most recent and therefore the most valid comparison models for the recognizer.
3. Instead of imposing fixed-size word-template directories in conventional files in computer memory, it is postulated that by using language-integrating programs such as (14), which look at cognitive associations between words, a better model of contextual word recognition will result. In concept, a relational database will be used, wherein the linking of templates will be dynamically determined, based upon the context of the task in progress. In effect, we will at a very basic level, be modelling human cognitive processes by passing a task-specific syntactic/contextual filter through the database, making and breaking linkages as appropriate (10). We believe that if these three limitations of current voice technology can be overcome, not only will voice recognition hardware be easier to use but a whole new world of computer-related devices will become accessible to a wider spectrum of disabled users.

CONCLUSION

This work does not set out to duplicate the considerable progress thus far achieved in the fields of artificial intelligence and data structure analysis (12). Rather we will be applying this knowledge to address the identified deficiencies of current voice input technology in the rehabilitative aid context. We will use the same Interactive Evaluation model used previously and will therefore be able to demonstrate any improvement in quantitative terms. We believe that before voice recognition can become an acceptable and widely used modality for human-machine interaction, the "personality" of the interface will have to be tailored to be more human-like. In short, future systems will need to have *sympathetic ears*.

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19.7

COMPUTER-AIDED MOTOR ASSESSMENT AND RATE PREDICTION FOR PRESCRIPTION OF COMMUNICATION DEVICES

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ABSTRACT

As part of the development of a systematic, computer-assisted approach to prescription of communication devices for the motor disabled non-vocal [1], the authors have designed a technique for estimating the communication rate a client will achieve with each of the devices being considered for him/her. Under the Prescription Procedure, predicted rate will be one of a small number of "match scores" which will be computed for the clinician as a means of rating the fit of each device to a client's abilities and needs.

CLINICAL PROBLEM

Other things being equal, clinicians would choose for a client the device providing the best communication rate. Even small differences, e.g. between 3 and 6 words/minute can make a major difference in the willingness of receivers to engage in dialog [2]. The problem is that communication rate is inherently a property of the person/device system; a device which is the fastest available, given one user's abilities, may be among the slowest for another user. The implication is that in order to know the rate at which a client will communicate using devices A through Z, they must be tried out with that client. The trial use approach, however, is flawed on both practical and theoretical grounds. Trial use of all devices -- even excluding those which could not be used by a client due to inadequate abilities -- is infeasible because there are simply too many. More importantly, the communication rate measured after one or two hours of use in the clinic may be completely unrepresentative of the "expert" rate the user will achieve after extensive use.

NEW METHODOLOGY FOR SOLUTION

What is needed is a means of determining a potential device user's expert rate as a prescription criterion, without the impractically lengthy practice period required to measure it directly with each candidate aid. The authors have developed a system of computer-assisted assessment and calculation procedures which meets this need. It consists of two major components. The first is an objective assessment technique which taps the motor abilities relevant to use of most devices. It is conducted using special-purpose testing instrumentation rather than any specific communication device. It is intended to bypass the cognitive demands which would be present in unpracticed use of a device, thereby measuring the motor abilities which set an upper limit on rate in expert use. The second component of the system is a computer program which combines the outcome of a client's motor assessment with information specific to the use of each candidate device to derive an estimate of communication rate. Since this estimate is based on data gathered in assessment designed to establish the upper limits imposed on rate by the client's motor abilities, it is referred to as Motor-Determined Maximum Rate, or MDM Rate. In effect, this scheme is intended to model client's performance and its variation with the physical characteristics of devices. It utilizes this model as a means of conducting, in the domain of computer-based estimation, the tests of expert device use which are impossible to perform literally.

MOTOR ASSESSMENT

The instrumented assessment scheme introduced above is intended, ultimately, to measure most motor abilities which could be harnessed to operate electronic communication devices. Attention to date has focused primarily on assessing the dynamics of movement among switches and their actuation. Three assessment methods, "Types 1 through 3", have been developed for assessing motor abilities relevant to the use -- respectively -- of keyboards or touch panels, for direct selection or encoding; scanners; and encoded devices with remote switches distributed among control sites on the user's body. The remainder of this article will be devoted to Type 1.

Type 1 assessment is based on the idea that the speed with which a user can activate the "keys" of a planar multiple-key interface, e.g. Canon or Autocom, will depend on a relatively small number of variables which define the motoric demand of the task. These may be expected to include distances of required movements, their direction and location with respect to the user's body, key size, and the force and travel required for key actuation. Type 1 assessment requires the client to perform a set of movements among targets, movements which span a manageable range of values for these task variables. The collected data is used, as described below, to model the dependence of the client's movement time on the values of these variables.

Because the intent is to establish the upper limit on motor performance, i.e. the minimum time for each set of values of the test variables, Type 1 testing takes the form of a series of "reciprocal tapping" tasks. The subject is asked to tap two targets, alternately, as rapidly as possible without making errors. Twenty taps, i.e. ten movement pairs, are required per task. The use of only two targets eliminates visual search and reduces cognitive demand. Inclusion of multiple taps in each task allows the collection of more data for a more reliable estimator.

The apparatus used for Type 1 assessment, for the case of zero switch force and travel, is easily portable and DC-powered. The tapping targets consist of brass disks of 1/2 inch, 1 inch, and 2 inch diameters. These targets are fixed in the positions required for each task by means of a square 3/8 plastic test panel - 28 inches on a side. Distributed on two radially symmetric patterns in the center and one corner quadrant of the panel are standard electrical "banana" jacks. On the back of each target disk is a mating banana plug which secures the target and allows electrical connection to the wiring on the back of the panel. The central pattern includes 20 pairs of target positions, requiring movements in the lateral, anterior-posterior, and diagonal directions (when the panel is aligned symmetrically in front of the client). Movement distances of 1-1/2, 3, 6, 12, and 24 inches are provided at each angle. An identical pattern of target positions, minus the 24-inch movements, is centered in one corner quadrant of the board. By rotation of the board, this corner can be located proximally or distally, right or left with respect to the client. Contact is sensed by means of a capacitive switch circuit (generously provided by George Kondraske of the University of Texas at Arlington) mounted on the back of the panel.

Data on movement timing is acquired by means of a Radio Shack Model 100 portable computer with 32 kilobytes of RAM. A special purpose interface has been built which accepts up to 8 mechanical or capacitive switch lines and also provides output to an opto-electronic scanning display used for Type 2 assessment. Physically, the interface is housed in a plastic enclosure roughly one inch high underneath the computer. It couples to the Model 100's bus expansion socket and interferes with none of the computer's standard functions. Data is acquired and stored under the control of a program written in BASIC. Data for each task, including experimental condition codes, requires a record length of 52 characters. This means that several clients' data may be stored in RAM (battery-backed) prior to transfer to disk files. Transfer is accomplished by means of a serial data line to an IBM Personal Computer, through a BASIC program running on that system. Note that each individual value for T is stored, not simply the average for a 20 movement task.

Which movement tasks are required of the client, i.e. the values of A (movement distance), W (target diameter), θ (direction) and Q (the quadrant of the test panel in which the movement is performed), defines the assessment protocol. This varies, to some extent, from client to client according to limits on range-of-motion and positioning accuracy. To date, Type 1 testing protocols have included as many as 192 tasks for able-bodied test subjects and as few as 32 tasks for disabled subjects with range of motion limited to 6 inch movements in one quadrant. The continued collection and analysis of data will be necessary to establish more definitively the set of tasks required to derive a reliable model of a client's motor performance.

RATE PREDICTION

In the present developmental phase, a subject's data files are analyzed by means of SYSTAT, a standard statistics package available for the IBM PC. In particular, the data file is sorted into subfiles, each of which contains the data for a subject for a single value of Q and θ. Each of these subfiles is then subjected to bivariate multiple regression, i.e. curve-fitting, in order to model the dependence of movement time on task variables and derive statistical measures of fit of the data to each model. In this way a separate set of model coefficients is obtained for each Q and θ. Four models are presently being utilized in this analysis to establish whether any is clearly superior. These are --

$$T = a + b \log_2 [2A/W] \text{ (i.e. Fitts' Law)};$$

$$T = a + b\,[A/W];$$

$$T = a + bA + cW;$$

and $$T = a + bA + c[1/W].$$

Calculation of MDM Rate is performed by a program written for the PC called DECOMP. Its use is straightforward. Prompting messages request entry of the model coefficients which have been determined for a particular client. A separate set of coefficient values may be entered for each movement direction, θ. DECOMP also queries the user as to the communication device for which a prediction of rate is to be calculated. Complete information on the language elements, the locations (x, y coordinates) of the keys by which they are selected, and the selection code, if any, is available to DECOMP on disk files. For standard devices, that information is entered only once in laboratory preparation of those files. In the case of custom or partly user-programmed aids, clinical entry of what the elements are and where they are located will be necessary. The other piece of information required by the program is the body of text it should use as a representative language sample. This can either be keyboard entered, or, as is the case in the present study, the contents of a previously prepared text file may be read.

The output of DECOMP -- which requires an execution time on the order of 2 minutes for alphabetic devices and a 550 word text sample -- includes the following:

C ≡ average number of device menu items required to form a word;

L ≡ average number of keyboard entries required to select a menu item (1 for direct selection devices);

T ≡ average time per keystroke;

LT = average time per character;

CLT = average time per word; and

1/CLT = average number of words per minute, i.e. MDM Rate.

DECOMP "decomposes" the text into the sequence of device units which would have to be chosen in order to transmit that text. It then further decomposes the unit string into the required sequence of keystrokes. (This operation has meaning only for encoded or "multi-level" devices for which each key strike selects a code entry rather than a language unit.) The decomposed "corpus" of text is next analyzed to determine the frequency of occurrence of each keystroke pair. The significance of a keystroke pair is that it determines a movement whose distance, direction and location are readily calculated from the device keyboard layout information which is initially read into DECOMP. Then, the client's model is applied for each movement to predict the time which would be required to accomplish it. The mean of these movement times, each weighted according to its frequency of occurrence, is T, average time per keystroke. From this, all the other predictions, including MDM Rate, are calculated.

PILOT RESULTS

The assessment and rate estimation methods described above have, at this writing, been developed to the level of working prototypes. Substantial human subject testing has been and continues to be conducted with disabled and able-bodied individuals. Fourteen motor-disabled adult subjects and ten able-bodied controls have participated to date. Etiologies include head injury, ALS, CP and stroke. Under present and future funding, more detailed experiments and further methodological refinements are anticipated. Nevertheless, present data supports certain tentative conclusions:

- Clinical feasibility is established. A protocol requiring 32 tasks has been accomplished by most of our motor-disabled subjects in periods under two hours. Setup time is minimal and subjects seem more intrigued by its novelty than bored by its repetitiveness.
- While learning or fatigue trends may be seen in the data for more than half of subjects, these are very small effects relative to variation of movement time with the task variables. This suggests that the effect on rate prediction accuracy introduced by time-varying performance will be minor.
- The models work. While a smaller proportion of the total variance in each disabled subject's data is explained by the regression equation than for abled-bodied subjects, not one of the present disabled subject group failed to show significant dependence of movement time on amplitude and target size. Often, correlations are surprisingly good. This statement holds for a headstick user and a mouthstick user as well as hand/finger users.
- Low speed is not necessarily coupled to high variance. Data from the slower subjects, or slower movement directions for a given subject, sometimes gave the best fit to regression equations.
- Differences among the models are not significant other than the expected improvement sometimes seen in correlation coefficient for the models having two separate independent variables.
- Movement direction can make dramatic differences. These differences show up most obviously in mean movement time. Even reciprocal movements which occur alternately in the same task can differ in the time they require by large percentages. The values of model coefficients, as well as the absolute and relative goodness of fit of the four models, also vary from θ to θ. This suggests the importance, in general, of allowing for angle-dependence of the model in using DECOMP to generate rate predictions.
- The few comparisons made to date between measured communication rate and MDM Rate reveal the expected qualitative relationship, with the latter exceeding the former by a factor of two or three.

In conclusion, it appears to be generally true that, while random variation is greater for motorimpaired than for able-bodied subjects, simple models succeed in extracting the reproducible average aspects of their performance. Reasonably reliable estimators of the dependence of movement time on task variables may be derived from clinically practical amounts of assessment data. This suggests that, while statistical validation of rate predictions remains to be accomplished, clinical feasibility of the investigators' methodology is confirmed.

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PACA: PORTABLE ANTICIPATORY COMMUNICATION AID

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ABSTRACT

A portable personal-computer-based communication aid is described which anticipates letter and word selections as a means of improving the efficiency of novel message generation. By this technique, the PACA, or Portable Anticipatory Communication Aid, enhances the traditional approach to scanning as a selection technique. The aid is configured on an Epson HX-20 computer. The dynamic selection arrays are presented on the Epson's built-in LCD display. In addition to prediction-assisted letter and word selection, encoded messages can be stored and retrieved.

INTRODUCTION

The purpose of a communication aid is to provide the user with an alternative mechanism to express thought. This is commonly done by mapping the language of the user, or, more accurately, linguistic elements derived from the language, onto a physical device. Direct selection letter/word/message boards and scanning arrays represent such mappings. This physical representation of language elements defines the language space of the communication aid (1). Communication is effected by constructing the expression from the available set of language elements.

Creating an appropriate language space requires a balance of articulateness, fluency, and efficiency. Articulateness refers to the exactness with which a thought can be expressed from the available language elements. Fluency is the rate at which an expression can be transferred to the recipient. And efficiency, by our definition, is a measure of the number of actions required to construct the message, whether they be physical actions made by the user of the aid, scanning actions performed by the device itself, or a combination of both.

Complete articulateness can be achieved by spelling any message letter-by-letter. However, to improve fluency, a communication aid should also incorporate a variety of more complex linguistic elements, such as words, phrases or whole messages. And the entire set of linguistic elements, the language space, should be organized efficiently to reduce the time required to construct a message and to reduce the number of individual items that have to be selected to make up the message.

A central problem in machine-augmented communication is the efficient and rapid production of novel messages. These types of messages account for the vast majority of communications (2). The magnitude of this problem is particularly apparent to users of scanning communication aids. The nature of the scanning technique, traditionally implemented with a static selection array, limits timely acquisition of the desired message elements. Although complete articulateness may be possible, fluency and efficiency are usually restricted.

Figure 1. Scanning Portable Anticipatory Communication Aid configured on the Epson HX-20 computer.

One approach to enhancing the production of novel messages with a scanning communication aid is to rearrange the contents of the selection array in anticipation of what the user may need to select. Some linguistic criteria is used to predict subsequent selections based on past selections.

BACKGROUND

Message Anticipation

Two basic schemes have been proposed which describe the manner in which the communication aid user interacts with the predictive process. In the first scheme (3,4,5), the user makes an initial selection and the aid makes a guess to complete the word or message fragment. The user then either rejects the guess or accepts it and continues with the next portion of the message. If the user rejects the guess, by making another selection, the aid attempts another guess, now with more information, i.e. two selections, and the user accepts or rejects it. With each additional selection, the prediction program attempts to make a guess (presumably with increasing probability of success) until a correct guess is made, or the possible guesses are exhausted, or the message item is completed by the user.

This process can be characterized as a form of "twenty questions". The user makes an initial choice. The aid responds by asking (figuratively), "Is this the rest of it?" If not, more information is provided by the user and the aid asks again, "Is this the rest of it?", and so on.

The accuracy of the predictions influences the rate and efficiency with which the user constructs the message. Each incorrect guess penalizes the user because the decision to reject it requires time and distracts the user from the desired content of the message. Prediction accuracy can be improved by "waiting" until the user has selected enough of the word or message that the program can make a guess of high probability of success. However, this approach may not bring about an improvement in rate (over a system which begins guessing with the first selection) because the user is required to make more selections before the system attempts a prediction. To optimize the interaction, one must counterbalance the number of user-made selections before a guess is attempted against the likelihood of an incorrect guess.

An alternative to prediction by a succession of single guesses is to have the communication aid offer a set of possible guesses after the user makes some initial selection(s) (3,6,7). By presenting several guesses simultaneously, the probability that the group contains the correct guess can be high even if the user has made only a few initial selections.

Rather than have to make a series of decisions to accept or reject guesses while trying to spell a word, the user can either select from the set of predictions or reject the entire set. The acceptance/rejection decision process is less fragmented and may, therefore, be less distracting. From the user's perspective, this form of anticipatory word completion is more akin to "multiple choice" than to "twenty questions".

A predictive scheme based on multiple letter and word anticipation had been developed at our center (6). Simulations of this scheme as well as tests with subjects demonstrated a significant reduction in the number of scanning steps needed to construct a message compared with the use of letter arrays alone. These results have already been applied to a scanning-based computer input device (8).

We have now implemented a variation of this predictive scheme on a portable scanning communication aid as a means of enhancing a person's ability to produce novel messages. The aid is called the PACA, an acronym for Portable Anticipatory Communication Aid.

IMPLEMENTATION

The utility of a communication aid is directly related to the number of communication modes which it can encompass. The PACA was proposed to augment person-to-person (or conversational) communication, note-taking, writing, and calculating capabilities. To achieve these functions, the device had to include: a dynamic display on which the selection arrays and word lists could be presented to the user, computational power and memory to handle the anticipatory algorithms and data base, temporary and permanent means of message output, portability, and at least full-day battery operation. Also, recognizing that hardware development costs passed on to the purchaser frequently place communication aids beyond the financial reach of their potential users, we proposed to develop the PACA around a commercially available portable computer which retailed for less than $1000.

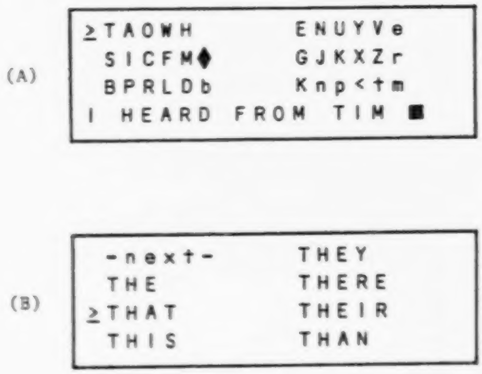

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Figure 2. Example scanning arrays of the PACA: (A) reordered letter array following the selection of a space and (B) frequency-ranked word list following the selection of "TH".

At the start of this project (July, 1984), the Epson HX-20 computer was the choice most compatible with our design criteria (see Figure 1). Its comparatively small LCD display was not considered a significant counter-recommendation because our approach to message anticipation required only small sets of highly probable choices to be presented.

Character Selection and Letter Anticipation

Characters are divided into two selection arrays: one for letters and one for numbers and symbols. Each array consists of six rows of characters divided into two three-row groups displayed side-by-side in the first three lines of the LCD display (see Figure 2-A). Each row contains characters for the construction of the message and symbolic characters (displayed as graphic symbols or lower case letters) representing functions (eg. backspace, erase message, print message).

In the letter array, the letters (displayed in upper case) are ordered by their probability of following the last selected letter (digram frequency) and the number of scanning steps needed to select an element from the array. Following the selection of each letter, the letter array is rearranged to place the most probable letters to follow the last selection in array locations that require the least scan steps to reach.

The message, as it is constructed, appears in the fourth line of the LCD display, beneath the character array. An individual message or message segment can be as long as 256 characters and can be reviewed at any time before it is output or stored. Longer messages of up to 508 characters can be constructed by retrieving prestored segments.

Word Anticipation

At the user's option, spelling by letter anticipation can be augmented with word anticipation. Following the selection of the initial two letters of a word, the letter array and message line are replaced by a list of probable words (see Figure 2-B). If the desired word is in the list, the user scans to the word and selects it (and a following space). If the word is not in the list, the user can examine additional lists, if more words are available, or scan past the word list(s) to return to the letter array to complete the spelling of the word.

Words appearing in the lists are taken from a set of core words, extracted from studies of large text samples, and learned words, initially spelled letter-by-letter by the PACA user during message creation. The words are presented and ordered by frequency ranking; core words first, then learned words. The core words provide a measure of stability to the presentation of common cross-message words while the learned words reflect the unique language needs of the user.

At the user's choice, a learned list can be saved and stored on the built-in datacassette of the Epson computer. At some later time, when the user returns to the topic under which a particular list was developed, the list can be recalled, biasing the rank-order word presentations in favor of the topical words.

Message Storage and Retrieval

Although pre-defined messages may be of little value in spontaneous or novel communications, they can be used to good effect in initiating communication, in providing feedback during conversation, in giving specific information, in asking questions, in making notes for personal use, and as routine fragments incorporated in otherwise novel messages.

Any message or sequence of characters appearing in the message line can be stored as a unit under a user-specified label. The format of the label is left to the user. Possible formats would include numerical codes, abbreviations, acronyms, or key words; whatever best served the user as a mnemonic aid.

To retrieve a message, the user first selects a "label-to-follow" symbol from the character array, then begins to spell the label. As the label is being spelled, the PACA program checks the message labels stored in memory. As soon as enough characters of the label are entered to unambiguously determine the completed label, the message is automatically retrieved and displayed in the message line.

SUMMARY

Rule-based anticipation, or prediction, of message elements can enhance the efficiency with which a scanning communication aid user can generate novel messages. The Portable Anticipatory Communication Aid, or PACA, is an implementation of this technique to aid in conversation, writing, and note-taking. Developed around the Epson HX-20 personal portable computer, the PACA also represents a cost-effective alternative to the traditional hardware design of communication aids.

ACKNOWLEDGEMENTS

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The authors also wish to acknowledge the contribution of Jerrilyn Voda, Senior Speech-Language Pathologist of the Rehabilitation Institute of Chicago, who conducted clinical trials of the PACA.

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A NEEDS-FEATURES SPREADSHEET FOR COMMUNICATION DEVICE PRESCRIPTION

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ABSTRACT

Lotus 1-2-3 spreadsheet software has been used as a structure for representing the communicative needs of non-speaking individuals and the ways in which these needs can be met by combinations of quantitatively measured features of non-vocal communication devices. This structure allows user-specific questionnaire responses, on the one hand, and files of data representing device features and functions, on the other, to be read in. These data are then processed according to formulae which represent the conditions for determining whether each need dealt with in the worksheet is present or absent for this individual and establishing whether each device in question possesses adequate levels of the relevant features to meet each of this individual's needs.

INTRODUCTION

Spoken communication becomes difficult and even impossible for substantial numbers of people with neurological conditions which lead to impaired motor function. The large number and diversity of technological aids and personal computers with specialized software and control interfaces, and the real differences among these devices in terms of potential benefit to the disabled individual, have focused increased attention on improving the process of prescribing communication aids.

A research project at Tufts-New England Medical Center and the Massachusetts Institute of Technology was funded by the National Institute of Neurological and Communicative Disorders and Stroke (NINCDS) in 1982 for this purpose[1]. As part of this research, a computer-aided system has been devised to guide the clinician in identifying the device or devices which seem to offer maximum communicative benefit for each particular client. This system uses client information in the form of answers to an extensive questionnaire concerning the client's needs and preferences; and device information, in the form of a lengthy set of quantitative laboratory measurements and ratings, stored in computer disk files. Another major aspect of this project, i.e. the prediction of communication rate, is dealt with in a companion paper [2].

The output of this system for each client-device pair is a 'Benefit' score, B. The system is intended to be used as a way of evaluating many devices for each client. The B score, then, represents the relative value of each device for that client in terms of predicted communicative benefit. In practice, the assessing clinician enters answers to questionnaire items provided by the client and others and the system prepares a B score for each device the client would be able to use.

The basic tool used by the investigators for representing and manipulating the complex, many-to-many "mapping" of client Needs to communication device Features is a commerciallyavailable software package known as Lotus 1-2-3. Development of this worksheet and its clinical application requires therefore the use of a computer which can run Lotus 1-2-3. We are using an IBM PC but the availability of Lotus 1-2-3 for other personal computers is expanding rapidly.

Lotus 1-2-3 was designed as a powerful "spreadsheet" for accounting purposes. It is essentially a very large grid of numbered rows and lettered columns, with the capacity to handle logical and arithmetic relations among its cells. In typical business applications, the user sets up mathematical relationships among "cells" of the grid and then observes the effect of entries in various cells which represent costs, income, tax rates, etc., on a particular cell, e.g. profit. At any time only a small portion of the grid is visible on the video monitor. The spreadsheet user has numerous means for travelling around the spreadsheet, including moving to a certain cell by entering the name with which she or he has labelled it, specifying a cell's column and row coordinates, or using cursor-control keys. If the user needs to look at two non-adjacent parts of a given row, a worksheet "window" function allows holding, e.g. column B on the left side of a split screen, and scrolling over to, e.g. column I on the right side of the screen.

SPREADSHEET STRUCTURE

The Needs-Features worksheet in its current prototype form makes use of 23 columns, A to W, each with a particular role in the calculation of the Benefit score. The need for so many columns stems primarily from the fact that conceptualizing the relationship between the client's needs, e.g. the need to communicate frequently with unfamiliar people; and such quantitative features as, for example, the number of characters in an opto-electronic display, is much more readily handled if intermediate levels such as 'Specifications' are posited between 'Needs' and 'Features'. Typically a Need is expressed in functional terms, the Specification names the device characteristics which in combination or alternatively can meet the Need, and the Ratings and Values columns which follow the Features list provide the measurements of the elementary features of each device which, in combination, meet or fail to meet each Specification.

Following are the headings of the Needs-Features worksheet columns as currently configured, with some illustrative example and explanation. Note that the Comment columns and some of the name columns which are included in the present version will be deleted in clinical use, with consequent savings in memory.

A - Question [Example: Do any people you deal with have difficulty seeing?]

B - Answers: [1 No
2 Yes - slightly
3 Yes - significantly
4 Yes - very much so
5 Yes - blind]

C, D, E, & F - Answer Entries: 1 or 0.

C - Answer for care giver(s).

D - Answer for family/significant other(s).

E - Answer for other people important to client, e.g. other friends.

F - Answer for other people with whom the client associates, e.g. fellow workers, and fellow residents.

G - Comment on Need (used in system development to clarify meaning, etc.).

H - Need (label). Each label in this column is the name assigned to the cell to its right.

I - Need (value: 1 signifies Need present; 0 signifies Need absent). Formulae are stored in this column which determine the presence or absence of a need (named in H) based on answer entries in C through F. A given Need formula is not displayed in the cell but can be called up for editing. The value computed by the formula is displayed in the cell. [Example: The Need for "Ambulatory Portability" depends on "yes" answer to a question about whether a client is capable of ambulation.]

J - Comment on Need Weight.

K - Need Weight (label).

L - Need Weight (value: any value from 0 through 1). The formulae in this column calculate the weight to be accorded each need on the basis of answers (C, D, E, & F) to questions. The weighting factor calculated for the client being evaluated is displayed. Some weighting factors are established directly via specific questions about strength of preference, while others are implied by answers to several questions. [Example: The weighting factor applied to "Ambulatory Portability" is a function of how much use the client makes of his or her walking ability.]

M - Benefit Increment (value: 0 through 1). This column divides the left side of the worksheet, which profiles the client, from the right side, which operates on device-specific information. The formulae in this column compare a Need value and the corresponding Need Met value for a device (see below). If both are 1, i.e. the client does have that need and this device does meet it, then the value computed by the formula is 1 multiplied by the corresponding weighting factor. A need not met, or a need met which the client doesn't have results in a 0 value. This value is displayed in the Δ B cell for that Need.

N - Comment on Need Met.

O - Need Met (label).

P - Need Met (value: 1 for Yes, 0 for No). The formula which determines what 'Specifications' must be met in order for a given Need to be met is stored here. The value computed by the formula (for a given user and device) is displayed. [Example: The formula which determines whether the Need for "Ambulatory Portability" is met by a given device depends on whether Specifications are met for small size and low weight and short battery charge time and long battery life and low number of components and availability of carry pack or clip.]

Q - Comment on Specification.

R - Specification (label).

S - Specification Met (value: 1 for Yes, 0 for No). This column contains the formulae which establish the threshold levels which must be met by each relevant feature in order for each specification to be met. [Example: The specification for small size is met if the longest dimension of any component is $\leq$ 6 inches and the shortest dimension is $\leq$ 2 inches.]

T - Features (label).

U and V - Feature Values:

U - Rating Value. Features whose values are listed here are unitless indices calculated from measurable parameters or judged subjectively. [Example: A "visual busyness" rating is assigned on the basis of density of display characters, color intensity and contrast, glare, and number of controls.]

V - Objective Parameter Values. Entries in this column represent an exhaustive numerical (and Yes/No) profile of a device.

W - Units. The units of the values in column V are specified here. Comments on ratings are also entered here as needed.

SYSTEM USE

A file is compiled for each patient whose communicative needs have been evaluated, consisting of the answers given by the patient and the clinician (columns C through F). Similarly, each device is fully specified in a device file (consisting of Features, Ratings and Values). Once a client file is read in by keyboard entry, individual device files are read into the right hand side of the worksheet, from disk. B is calculated for each device as the sum of the B increments. Determination of Needs is structured so that the client's needs may be calculated independently of device information (i.e. Needs are conditional only on Answer values or other Needs).

The major advantage of this application of the Lotus 1-2-3 spreadsheet program is that it permits systematic, comprehensive, individually-tailored evaluation of the potential benefit each existing device may have to offer a client. Hand calculation of such a score would be virtually impossible. Another advantage is the capability it will offer the clinician to review the basis for a particular Benefit rating by examining what underlies it. An overwhelming advantage for the developers of the Prescription Procedure has been the facility with which the numerous, criss-crossing relationships among the client's needs and the features of devices can be represented and modified. It is like having the help of a thoroughly dependable colleague who is awfully touchy about protocol but has an enormous capacity for retaining detail and complexity. It contributes no wisdom or judgement but does encourage clear thinking by demanding precision of logic and consistency of format.

Currently the worksheet is being tested and adjusted using questionnaire information from 20 non-speaking patients who use communication devices or boards. Measurements of the features of the communication boards and electronic devices used by these patients, in addition to the other devices which we have evaluated so far, constitute the device data files. An experimental study which will test the predictive value of the Needs-Features Spreadsheet along with other aspects of the Tufts-MIT Prescription Procedure is scheduled to begin in April, 1985.

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BEST COPY AVAILABLE

INSTRUMENTED, ADUSTABLE SEAT FOR EVALUATING POSTURE AND BODY CONTOURS

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MCLAURIN, C. A., Sc.D.

ABSTRACT

A prototype of an adjustable fitting seat was designed and built to aid in body positioning and recording of anthropometric data. The seat incorporates hydraulic jacks for positioning and computer interfaced contour and pressure distribution gauges and rapid collection of data. The seat was evaluated with an anthropometric dummy and by a non-handicapped client. It will be used for research on modular seats for wheelchair use.

INTRODUCTION

The health and comfort of wheelchair users are dependent in a large part upon matching the seat to the user. Currently, the matching is accomplished by the clinic team using trial and error techniques. To aid in this process and to systematize data collection relative to seating an adjustable fitting seat was developed. Criteria for the design was the determination of optimum seating configuration based on clinical judgement, the recording of back and seat contours, to aid in contoured cushion design and the provision of disability specific anthropometric data for modular seat construction and the determination of the center of gravity of the subject.

The purpose of this paper is to describe the design of the adjustable fitting seat and to report the initial results. The location of the vertical line through the body center of gravity was also determined using an anthropometric dummy.

METHODS

The adjustable fitting seat has four support components plus the electronic measuring system. The supports are the three legged frame, footrests, hinged seat and the back support. The electronic measuring system consist of the load cells under each leg for body weight and center of gravity determination, two sets of contour gauges, and electronic goniometers for direct measurement of seat and back angles. The lateral view of the assembly is shown on Figure 1. One set of contour gauges is used to measure seat contour at the buttocks and the other is to measure contours of the back. The frame of the seat and the back are supported and adjusted by 1500 lbs. capacity hydraulic jacks for rapid adjustment and easy positioning. The seat is hinged near the front edge such that retracting the jack will increase seat angle and reduce elevation of the back edge of the seat. Seat depth is adjusted by fore and aft movement of the back support while second hydraulic jack will adjust the angle of he back. Leg length, elbow height, knee, foot and forearm angles are all manually adjusted by slide and wing nut mechanism. In Table 1 the range of available adjustments are listed relative to the horizontal plane and the front edge of the seat.

FITTING SEAT ADJUSTMENTS				
Seat Angle	0	to	20	degrees
Back Angle	90	to	135	degrees
Seat Depth	22.2	to	43.8	cm
Leg Angle	90	to	160	degrees
Leg Length	12.7	to	63.5	cm
Foot Angle	-38	to	12	degrees
Forearm Angle	-20	to	20	degrees
Elbow Height	25.4	to	34.9	cm

Table 1

Adjustments and their range of motion for fitting seat; all angular measurements are relative to the horizontal.

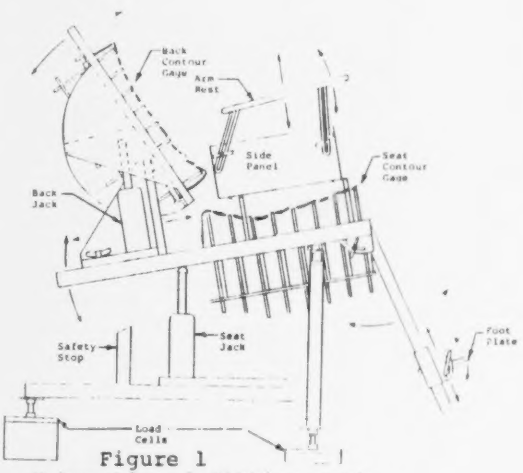

Figure 1
Schematic of fitting seat showing adjustments.

The contour gauges (1) are modular construction and utilize linear potentiometers. 64 probes extend through predrilled holes in the wheelchair cushion to the buttocks and are attached to the sliding potentiometers below the seat. Motion of the probe causes a change in the potentiometer voltage which is used to measure changes in contour at the body-cushion interface. Knowing the location of the potentiometers in the plane of the seat or back, the three dimensional coordinates of discrete points of the surfaces can be described. The contour gauges at the seat and the back each measure 16 x 16 inches area and can accommodate cushions up to four inches thick; with 64 predrilled holes for the sensor probes. All data points can be rapidly recorded by multiplexed input to A/D converters on an existing LSI-11/23 computer.

Simultaneously with the contour measurements the distribution of pressure on the buttocks is also obtained using various pressure sensors. The pressure evaluation pad developed at the Texas Institute of Rehabilitation Research (2) was found to be practical when interfaced with the computer. Using this system, pressures up to 100 mm Hg could be recorded in 144 locations under the buttocks in four minutes or less.

With the patient seated, the fitting seat is adjusted to a position judged appropriate by clinical experience and subject response. This can be verified by optimum buttock pressure distribution. The contours of the back and buttocks then recorded by the contour gauges. The load cell (3) output is sampled and the center of gravity for the patient is calculated. The weight of the seat components are removed in the calculations. The measurements are completed with the recording of body dimensions for off line correlation with the other data. This anthropometric data is collected to aid in the description of posture, specific to the disability.

RESULTS

The contour of the back and buttocks were found on a normal male sitting in the upright position with the seat at 0° and the back at 90° to the horizontal. The contours were defined at eight transverse sections 50 mm apart down the back from the inferior angle of the scapula and eight more transverse sections across the buttocks. Three representative samples of the back contours are shown on Figure 2 taken from various heights below the top edge of medium density polyurethane foam back cushion.

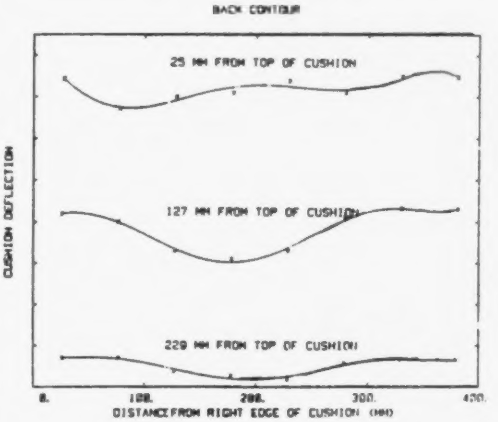

Figure 2
Back contours at various heights below scapula. Normal adult male supported by medium density polyurethane foam.

On Figure 3 representative samples of the buttock contours are shown for another polyurethane foam cushion (E & J Dura-Foam). The curves were fitted with a 6th order polynomial to smooth the discrete data and produced a standard deviation of ± 1mm of deflection of the tissue-cushion interface.

The contours showed smooth interfaces with double peaked surfaces at the seat and the back. Highest indentation occurred near the ischial tuberacities and at the low thoracic region at the midline.

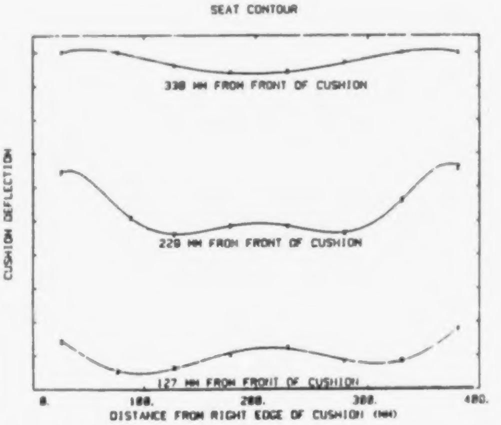

Figure 3
Seat contous of a normal adult male at different sections behind front edge of polyurethane seat cushion.

To test the accuracy of the center of gravity determination the Bratgaard dummy (4) was used. This body analog represents a 50 percentile human in the sitting position with accurately known center of gravity, mass and body curvatures. Repeated determination yielded a discrepancy less than 5% between the actual and measured location of the center of gravity in the plane of the seat. The height of the center of gravity above the seat was not needed and was not determined.

CONCLUSIONS

The adjustable seat was found to be easy to use and accurate for the measurement of disability specific anthropometric data, body contours and center of gravity location in the plane of the seat. The result will be useful to characterize clinically determined seating configuration for wheelchair applications and will provide engineering design data relative to body contour, pressure distribution and posture.

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20.2

THE RELATIONSHIP BETWEEN POSTURE AND ISCHIAL PRESSURE FOR THE HIGH RISK POPULATION

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ABSTRACT

This study was designed to determine the relationship between posture change and pressure redistribution on five wheelchair cushions or sitting surfaces. The new Oxford Pressure Monitor was used with one normal subject. It was found that the pressure relief characteristics of the various cushions is quite different and in some cases possess characteristics that are contrary to contemporary thinking.

INTRODUCTION

Research and clinical findings have indicated that the development of decubitus ulcers (pressure sores) for people with loss of sensation are unlikely if pressures above 35 mm Hg are relieved every 20-30 minutes (Kosiak 1959; 1961). Kosiak also demonstrated that pressures of 70 mm Hg for two hours could cause pathological changes in muscle tissues of rats and pressures of 60 mm Hg of one hour duration will cause changes in tissue of dogs. Mooney et. al (1971) and reported findings of ischial tuberosity pressures in ranges from 41 to 86 mm Hg on various seat cushions. Reswick (1975) also reported studies which suggested thresholds of pressure versus time related to specific anatomical sites. As a result of this and other related studies various regimes of pressure relief are currently practiced. More recent findings have shown that many paraplegics and tetraplegics can remain free of pressure sores with average ischial pressures considerably above the 35 mm Hg threshold level, for times between pressure relief greatly in access of the usually suggested care regime, Ferguson-Pell (1980), Fisher (1983), and Perkash (1984). Zackarkow (1984) through literature review and clinical observation has suggested that postures imposed by the standard wheelchair, combined with asymmetrical body alignment, have a significant bearing on a persons predisposition to pressure sores.

The design of commercial support surfaces (cushions) has been guided largely by these research and clinical findings. Essentially two directions have been taken: a) equalization of support pressures and b) redistribution of supporting pressures. The fluid-based designs (air, gel, water) attempt to equalize the pressure across the support surfaces. Pre-contoured support structures (VASO-P, A.S.S., and cut-out foam) basically attempt to redistribute the pressure away from high risk anatomical sites to areas more able to accommodate higher pressures. Custom contoured foam (Jay, Foam-In-Place) attempts to both equalize and redistribute support pressures.

It is generally accepted that paraplegics with symmetrical body alignment using one of the more advanced cushion designs, and who follow a reasonably conservative care regime, can expect to remain free of decubitus ulcers. However, individuals that have either asymmetrical body alignment due to deformity and/or pathological process or are unable to perform periodic pressure relief are still at high risk. For example, tetraplegics are encouraged to illicit posture changes either voluntarily or with the aid of powered recline mechanisms on their wheelchairs in order to achieve periodic pressure redistribution. The effectiveness of the various commercial cushion systems to accommodate the stability, pressure relief and postural support needs of this later group has not been as carefully studied.

The purpose of this study is to examine the relative effectiveness of the various generic types of support devices to meet the postural and pressure redistribution needs of this high risk population. This paper reports on the methodology and the results of Phase I; which is the normative data derived from a single normal subject placed in defined seated postures on five different support surfaces.

METHODS AND MATERIALS

The postures examined were anterior and lateral trunk flexion and trunk recline; within the maximum ranges that could be expected of a person while sitting in a standard wheelchair. The posture imposed by most standard sling type wheelchairs was used to define the neutral reference posture. This neutral posture is a midline orientation in the coronal plane with approximately 4° hip flexion and 6° of trunk recline (hip extension) from the vertical. Footrest loadings were adjusted to 12% of body weight (20 lbs. - 9 kgs for the test subject). From the neutral posture, three positions of anterior trunk flexion (15°, 25°, and 35°), two positions of lateral flexion (10°, 20°) and three positions of recline (20°, 40° and 50°) were studied. The support surfaces studied were: a) flat firm surface; b) VASIO-P; c) Roho cushion; d) Jay cushion; and e) Foam-In-Place System (FIP). The flat firm surface (1/2" plywood board) was used to provide reference data. The four cushions were chosen sincethey contained the most current developments that are commercially available. They also are representative of the various generic approaches to pressure sore management. Several additional cushions are scheduled for inclusion in the study Polyaire (Scimedics) and the Active Support Seat (Beneficial Designs).

The instrumentation chosen was the Oxford Pressure Monitor. This is a new instrument developed by the Nuffield Orthopaedic Bioengineering Centre. It features a pneumatic cell transducer matrix configured in 3 x 4 arrays, or 2 - 1 x 6 arrays. The transducer cells do not have electrical contacts as in earlier designs, but rather monitor changes in the pressure gradient. The transducer arrays were taped directly to the ischial, sacrococcygeal and posterior thigh areas of the subject (Figure 1). All readings were taken in a standard Everest and Jennings' adult size wheelchair equipped with a MED - type power recliner mechanism. Posture changes were measured by "devil levels" secured to a trunk harness by means of a posterior upright. Footrests were adjusted to attain the same foot plate loading 9 kgs (20 lbs.) in the neutral posture prior to each set of pressure recordings. Pressure recordings were taken in all 36 cells in each of the nine positions, for each of the five support surfaces studied. The test subject weighs 76.2 kgs (168 lbs.) is 171.4 cm tall (5'8") and is of muscular build. No body asymmetry of any nature was observed. Pelvic symmetry in the neutral position was obtained through observation and feeling of "equal support" as determined by the test subject. Calibration checks of the instrumentation were done both before and after each recording session.

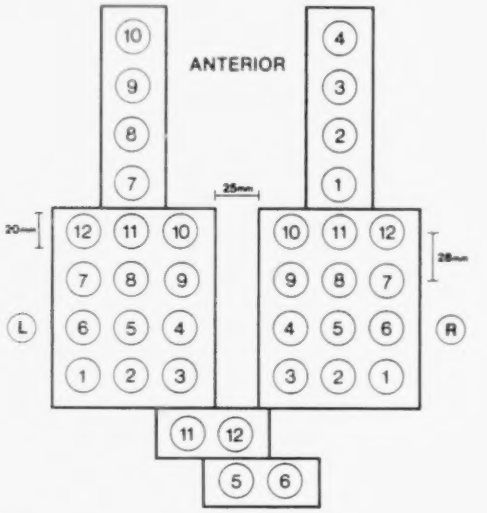

Fig. 1

RESULTS

Two plots of peak pressures are given in Figures 1 and 2. The first (higher value) is the hightest pressure recorded regardless of cell location. The second (lower value) is the pressure value of the cell determined to be at the location of highest risk. The cell location of the highest risk area was determined by noting where the peak pressure occured on the ischial transducer matrix (left or right) while seated in the neutral posture. The rationale being that this is the posture maintained for the most time when seated in a wheelchair and therefore the location at which ischaemic producing forces are most likely to occur. Plots showing only a single value simply means that the highest value and the high risk area value are the same and occur at the same site. The following are specific findings in terms of peak pressures with emphasis on the high risk area results.

Neutral Posture (Figures 2 & 3) Pressure measured at high risk areas exceeded 60 mm Hg for all cushions (VASIO-P 80, Roho 61, Jay Cushion 67, and FIP 62). The Peak pressures on the Roho and VASIO-P cushions occured at a location other than the high risk area (pre-ischial/ posterior thigh). The peak pressure of the VASIO-P is equivalent to that measured on a flat board (147 mm Hg). Significant differences between left and right peak values were meassured: VASIO-P: 49 mm Hg, Roho: 16 mm Hg, Jay: 5 mm Hg, and FIP: 11 mm Hg.

Anterior Trunk Flexion (15°, 25°, 35°- Figure 2) In general, forward flexion of the trunk caused increases rather than decreases to the pressures at the high risk sites: maximum changes, VASIO-P (+11%), Roho (+18%), and Jay (+6%). Decreases did not occur until 35° flexion and only on the Jay (-9%) and FIP (-8%). Peak pressures also showed a similar pattern. Foot plate loading increased 40-45% at maximum forward flexion (35°).

Lateral Bending (10°, 20° - Figure 2) Side bending to the left of 10° and 20° produced the following percent reductions to the pressures (mm Hg) at the high risk site on the right side; VASIO-P 10° -5%, 20° -6%, Roho 10° -28%, 20° -48%, Jay 10° -4%, 20° -19%, and FIP 10° -14%, 20° -29%. The minimum pressure values were obtained on the right high risk site during left trunk bending to 20°: VASIO-P: 75 mm Hg, Roho: 26 mm Hg, Jay: 46 mm Hg and FIP: 36 mm Hg. Peak values on the left high risk site at 20° left bend: VASIO-P: 93 mm Hg, Roho: 84 mm Hg, Jay: 66 mm Hg, and FIP: 46 mm Hg.

Recline (20° - Figure 2) Recline of the trunk (hip flexion) to 20° from the vertical caused the following percentage pressure changes in the highest risk areas (L or R): VASIO-P: +10%, Roho: -28%, Jay Cushion: +7%, and FIP: -27%.

DISCUSSION

The results suggest that the Roho, Jay, and Foam-In-Place all provide similar capabilities to distribute pressures throughout the support surfaces for a person seated in a neutral posture. Compared to the neutral posture, forward flexion up to 25° caused increases or no change to high risk area pressures. The highest percentage increase being Roho (+18%) and lowest FIP (no change). Thirty five degrees forward flexion produced decreases in only the Jay (-9%) and FIP (-8%) cushions. Side bending (10°, 20°) results show a wide range of non-linear pressure reductions to the high risk area on the contralateral side; largest Roho (-28% and -48%), lowest VASIO-P (-5% and -6%).

Assuming the 35 mm Hg is the threshold below which capillary flow is maintained, a dependant person may need to be reclined beyond 40° from

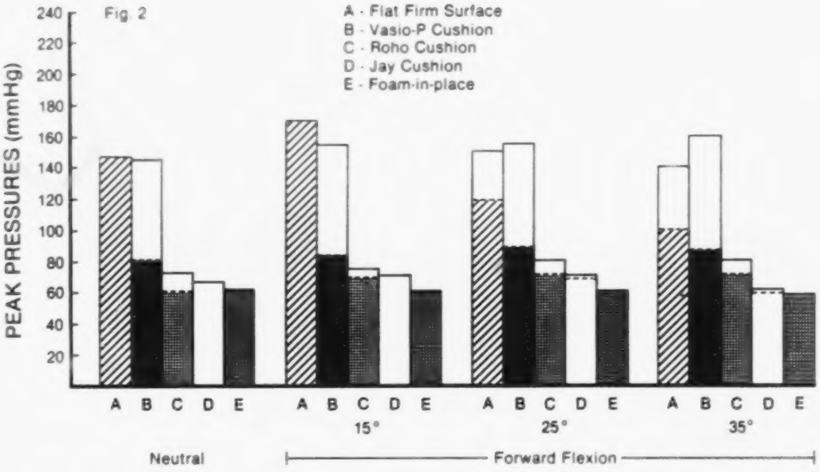
Fig. 2

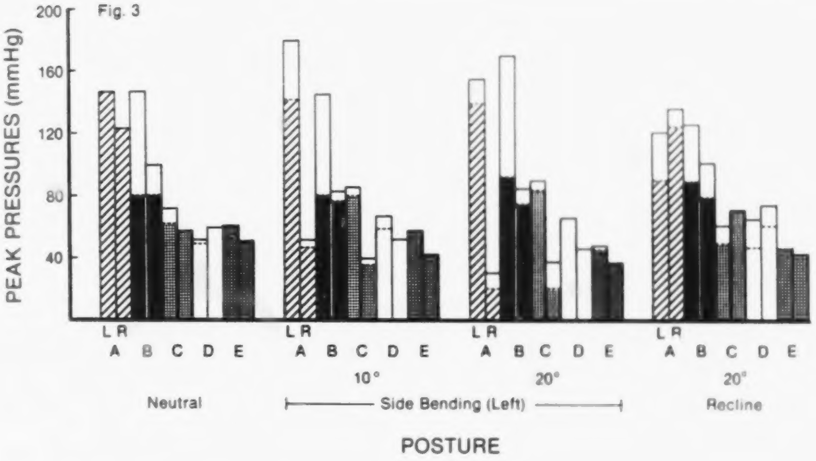
Fig. 3

the vertical to achieve this lower threshold value at the highest risk site. Peak sacrococcygeal pressures were similar for all cushions (30 - 40 mm Hg) except the VASIO-P which was significantly lower (20 mm Hg). The wheelchair backrest as well as the cushions themselves are both likely contributors to pressure in this area. And finally, it is recommended that pressure recordings be taken in both the neutral posture as well as those postures that can be assumed by an individual seeking partial pressure relief while seated in a wheelchair.

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SHAPE AND PRESSURE DISTRIBUTION ON WHEELCHAIR CUSHIONS

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ABSTRACT

Instrumentation has been constructed to simultaneously record interface pressure and contour plots in weight bearing conditions on wheelchair cushions. The data indicated significant differences between normal and flaccid paraplegic buttocks and between foam and dry floatation cushions. The results can be useful in the fitting of body supports for wheelchair mobility.

INTRODUCTION

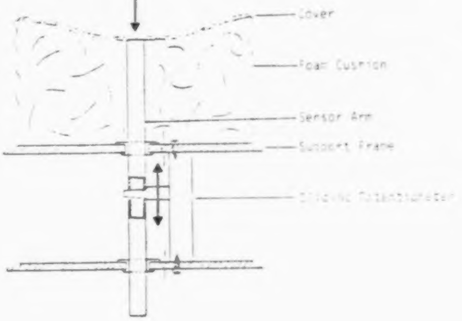
Under externally applied loads the mechanical properties of the tissues have an important role in determining the perfusion and nourishment of the cells in the weight bearing area. Load distribution and soft tissue deformation are two important properties controlling tissue survival in wheelchair sitting. Simultaneous measurement of pressure distribution and change in shape of the tissue interface were used to indicate the load and deformation of the tissues of subjects sitting on wheelchair cushions.(1)

In the present work, pressure and tissue contour plots for normal and paraplegic buttocks were constructed to aid in the analysis of tissue interface and to aid in the fitting of viscoelastic, polyurethane and dry flotation cushions. Successive recording of the contour plots also allowed measurement of load bearing soft tissue deformation by observing the development of the final shape of the tissue-cushion interface during load transfer to full support.

METHODS

A prototype instrument was built consisting of a modular contour gauge and the pressure evaluation pad, originally designed at TIRR, (2) and now interfaced with the LSI-11/23 computer. The contour gauge shown was constructed to measure tissue deformation using inexpensive linear potentiometers. In each module, eight potentiometers (Mouser, 100K ohm, 2.3 inches travel, $0.99/ea.) were fastened between top and bottom support plates and the slider connected to a 7 mm diameter aluminum tube probe. The probes, shown schematically in Figure 1, extended through predrilled 9 mm diameter holes in the cushion, are attached to the cover and transmit vertical cushion displacement to slider motion on the potentiometers. Motion of the probe causes a change in voltage which is used to measure changes in tissue contour at the weight bearing region of the buttocks.

Fig. 1



Eight modules, 4 x 8 inches each, have been assembled to form a 16 x 16 inch seat surface with sixty-four potentiometers. Knowing the location of the potentiometers in the plane of the seat, the three dimensional coordinates of sixty-four discrete points of the sitting surface can be described. No major change in support properties of the cushions was anticipated by drilling 9 mm diameter holes on 2 inch centers, since only 2.5% of the cushion material was removed and the support behavior of the cushion was altered minimally.

The linear sensitivity and reproducibility of the contour gauge have been characterized by the calibration curves of three potentiometers. All potentiometers were found to have a zero offset of less than 6 mm and a 45 mm linear displacement region. The offset was eliminated by a mechanical stop and the potentiometers were found to have the sensitivity of 9.61 ± 0.46 mm/V. The input was multiplexed from 64 to 4 channels into 12 bit A/D converters, with the sampling frequency of 3.3 KHz 20 msec per 64 samples was required. At the end, data display and printing time occurred in less than one minute.

Calibration of the pressure distribution sensor was made using plane, point and curved surface loading. The results were satisfactory but accuracy was dependent on the method of loading. The point load calibration showed the least accuracy but the accuracy on a wheelchair cushion with curved surface loading was better resulting in a discrepancy of 21 percent between applied and measured loads. The most accuracy was seen with flat, plane loading producing an accuracy better than 1 percent. In spite of these variations in accuracy, with methods of loading the calibration measurements could be reproduced to ± 1 mm Hg pressure.

RESULTS

Using this new instrumentation preliminary observations have been made to compare tissue deformation on three types of cushions for three types of subjects. The wheelchair cushions investigated were a three inch thick viscoelastic foam, a three inch thick polyurethane foam (Everest and Jennings Dura-foam) and a four inch thick ROHO cushion (specially prepared and supplied by the ROHO Research and Development Co., Inc., E. St. Louis, IL). All cushions were prepared with sixty-four holes for the sensor probes. The subjects were a normal male and female and flaccid paraplegic males.

Initial experiments using the tissue contour gauge were to study the effect of curve fitting techniques on the analyzed data. Using curve fitting the number of data points needed to define the interface could be reduced and thus fewer drilled holes were required in the cushions with the result in less alteration of support properties of the wheelchair cushions.

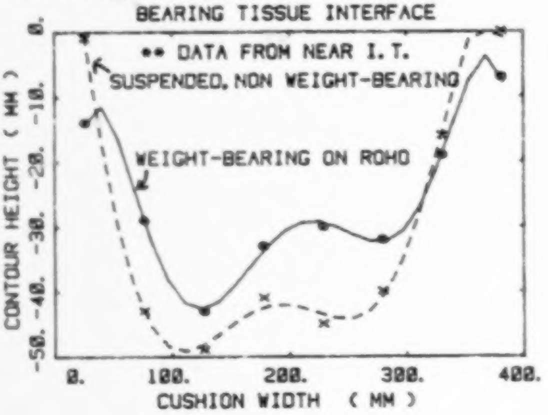
Eight discreet data points generated from each of several rows of the tissue contour gauge were fitted with increasing order polynomials from one through seven. The 7th order polynomial fit was inconsistent with the geometric shape of the tissue-cushion interface. The 6th order fit was highly consistent and gave ± 1.08 mm standard deviation of the data about the approximation polynomial Thus the 6th order polynomial was chosen to fit all contour data.

Data from the entire sitting surface was analyzed and rows and columns of sensor outputs were plotted for coronal and sagittal planes of the buttocks. Thus three dimensional models of the entire tissue cushion interface were constructed in some cases. To simplify the presentation in this report only transverse coronal plane data will be shown taken through or at the ischial tuberosities of each subject.

Development of Weight-Bearing Equilibrium Support Interface

The formation of the weight bearing tissue cushion interface was recorded in comparison to the non-weight bearing tissue contour near the ischial tuberacities. Initially the buttock contours were recorded with the subject suspended over the contour gauge without loading the skin or the sensor surface. The elongated, hanging tissue shape is shown on Figure 2. Latter the subject-transferred his weight to the ROHO cushion on the contour gauge and the compressed tissue shape was also recorded and shown on Figure 2. Note the absence of lateral displacement in this normal weight bearing data.

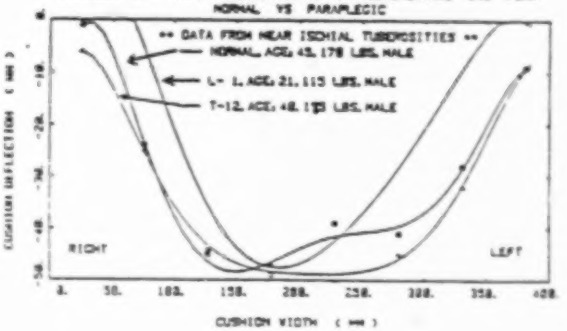
Fig.2 DEVELOPMENT OF NORMAL WEIGHT-BEARING TISSUE INTERFACE



Normal and Paraplegic Contours

Two flaccid paraplegic males were studied and their tissue contours compared to a normal male subject (see Figure 3). Both paralytics lacked muscle tone and have been paralyzed for three and four years. Characteristically a single peaked surface was seen for the paraplegics on either polyurethane or viscoelastic foam with peak deformation near the midline. The normal subject showed double peaked surface following the contour of the ischial tuberosities.

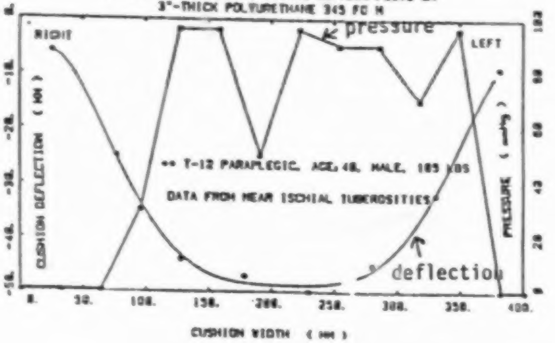
Fig. 3



Pressure Distribution Measurements
Pressure measurements were also studied on the two types of foam and on the dry flotation (ROHO) cushion. Interface contour and pressure were superimposed on the same plot as shown below. There was no attempt to do curve fitting on the pressure data. In general the results show that the profiles of pressure and contour are mirror images of each other and appear to be inversely proportioned for the foam cushions. The inverse relationship did not hold for the ROHO cushion, where the contour and pressure of the interface were directly related. Qualitatively the results for the ROHO cushion remained similar for inflation pressures below 60mm Hg.

The inverse relationship on polyurethane foam was also observed for a paraplegic subject as shown in Figure 4. The pressure increased to a high plateau across the central, high portion of the deflection area with slight decreases off center of the expected locations of the ischial tuberosities.

Fig. 4



CONCLUSIONS

The contour gauge was found to be useful in the measurement of tissue-cushion interface at 50 mm intervals throughout the buttock surface. The measurements allowed the observation of the development of the equilibrium interface and indicated the extent and rate of tissue deformation. The instrument used in the Seating Clinic will be useful in short and long term sitting studies. Coupled with temperature and pressure measurements it will provide a more complete description of the tissue-cushion interface.

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20.4

MAPPING THE DEFORMATION AND PRESSURE CONTOURS AT THE BUTTOCK-CUSHION INTERFACE

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ABSTRACT

A system for the simultaneous measurement of deformation and pressure contours at the buttock-cushion interface has been described. The interface contour is measured on a standard gel pad using an ultrasonic dimension gauge, and pressure contours are measured using a pressure evaluation pad. It is hoped that characterization of each patient by the measurement of pressure and interface contours on a standard seating surface will aid in the prescription of optimal seating surfaces for spinal cord injured persons.

INTRODUCTION

Clinical problems associated with the management of patients susceptible to pressure sores present a major challenge for the rehabilitation team. Spinal cord injured persons are at a particularly high risk for tissue breakdown. Current practice for their prevention consists of first identifying areas of high pressure between the body and the support surface and then selecting or fabricating cushioning that will redistribute the forces over a larger surface area and thereby reduce pressure peaks and gradients at the interface. However, the capacity to distribute the pressure uniformly at the support interface does not depend only on cushion properties; rather, it is clear that body build, weight, height, size and shape of bony prominences, and tissue resiliency are some of the factors that govern pressure distribution at the buttock-cushion interface. Garber and Krouskop (1982) have shown that thin patients, as a group, showed significantly higher pressure over the bony prominences than did average weight or obese patients. Further, these areas of higher pressures in the latter group occured in the regions of soft tissue rather than under bony regions. This is surprising in view of the greater load presented to the cushions by the average weight and obese patients.

Presently, measurements of pressures at the interface between cushion and skin are routinely used as an indirect indication of local tissue deformation in the selection and prescription of wheelchair cushioning in the clinic. There are, however, major limitations to the accuracy of these pressure measurements, especially when subtle differences in interface pressures are required to be detected or when absolute rather than relative values are needed. These would be typical problems encountered when working with persons with asymmetrical sitting habits, such as scoliotic patients. Therefore, it is apparent that more than peak pressures under the bony prominences have to be considered. Specifically, a cushion indentation contour, combined with pressure measurements, may be a more direct method of attacking the problem of prevention of tissue damage; data on the latter factor has not been available until recently.

TWO DIMENSIONAL SYSTEM

In 1983, a method for measuring the two-dimensional contour of the buttock-cushion interface using an ultrasonic dimension gauging technique was developed at Helen Hayes Hospital and demonstrated on three normal subjects for the first time (Kadaba, et al, 1984). The technique is based on the fact that an ultrasonic pulse traveling in a multilayered medium is reflected at each interface between two layers with dissimilar acoustic properties. Using the temporal locations of reflected echoes from the buttock-cushion interface with reference to an excitation pulse, it is possible to measure the deformation contour of the gel surface while a subject is seated on it.

The two-dimensional measurement system consisted of a modified wheelchair, with the seating surface replaced by a thick plexiglass plate with a transverse slot to facilitate accurate positioning of the ultrasonic probe. A thin plexiglass sheet was placed over the slotted plate to support the seating surface. A 3 cm thick polyvinyl chloride gel cushion was employed to provide a standard support surface on which to characterize the behavior of the subjects' soft tissue and bony structures under load. The cushion was acoustically coupled to the plexiglass sheet using a water based agent. The gel cushion was chosen as the standard because ultrasound does not propagate readily through foam materials.

To obtain the two dimensional contour, the subject was seated on the PVC gel, care being taken to align the ischial tuberosities with the slot in the plexiglass frame. The ultrasonic transducer was then manually positioned within the slot and acoustically coupled to the bottom of the thin plexiglass sheet. The transducer was then manually moved along the slot in 0.5 cm. steps, and the echo originating from the buttock-gel interface was digitized and stored for further analysis. The calculated buttock-gel interface distances and the transducer position are used to generate the contours. In these tests, it was noted that the three subjects caused three different indentation patterns on the same standard gel pad. Although the causative factors related to these different patterns were not clear, the shapes were obviously related to the manner in which the subjects' tissues transmitted loads.

As an extension of the two-dimensional system, an ultrasonic scanning system capable of recording the 3-dimensional buttock-cushion interface and pressure contours while supported by gel pads is currently being developed. The design concept includes the implementation of precision stepper motors to control the spatial location of an ultrasonic probe in an X-Y plane as well as the development of software responsible for data acquisition, analysis, and display. Pressure contours are measured using a TIRR Pressure Evaluation Pad interfaced to a computer. The interface contour, combined with the corresponding pressure measurements, will be of considerable value in the selection of cushion materials and in fitting wheelchair cushions for the spinal cord injured. Direct comparisons between contours on gel and foam cushions will be made using a probe technique (similar to that being developed by Reger, 1984) to determine contours on foam.

of binary values (S. Reger, 1984). The sensing pad is placed between the subject and the gel pad. The pneumatic system is activated by an air pump, and the pressure is monitored using a pressure gauge. By varying the air pressure within the pad and observing the matrix display on the computer, it is possible to obtain the pressure contours under the buttocks and upper thighs while the subject is sitting on a cushion. By locating the pressure pad accurately, it is possible to obtain values of pressure under bony prominences as well as under regions of soft tissue.

CONCLUSION

Within the clinic, problems are frequently encountered where pressure measurements and subjective evaluations alone do not provide sufficient information for prescribing appropriate wheelchair cushioning. Specifically, problems exist in assessing the degree of asymmetric loading in scoliotic patients and in identifying the highly localized pressure distributions of patients with flaccid tissue covering bony prominences. The aim of this study is to characterize an individual's interaction with a standardized seating surface and to use this information to determine whether or not an individual who produces substantial indentation of the seating surface may be considered at higher risk for developing decubiti than another of similar or greater weight who produces little indentation.

Therefore, a system for mapping the three-dimensional deformation and pressure contours at the buttock-cushion interface has been developed. An ultrasonic dimension gauging system, with a gel cushion as the standard seating surface, will provide the high resolution necessary to accurately map the deformation contour, while a pressure evaluation pad will provide the corresponding pressure contours. Initially, the device will be used to evaluate normal subjects of varying somatotypes; however, clinical evaluations will also be performed using both paraplegic and quadriplegic patients. Thus, differences between normal, paraplegic, and quadriplegic persons will be described quantitatively for the first time in terms of tissue under load.

By combining the mapping of a buttock-cushion interface contour with the corresponding pressure contour on a standard support surface it will be possible to provide more meaningful information to the clinician so that appropriate cushioning can be selected with reduction of trial and error procedures.

ACKNOWLEDGEMENT

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THREE DIMENSIONAL CONTOUR

A special chair was designed and fabricated using 2 cm O.D. stainless steel tubing. The back, arm and foot rests are similar to those on a standard wheelchair. The seat is replaced by a thick plexiglass sheet (2 cm thick) with a 30 cm x 30 cm hole cut in the center. A thin sheet of plexiglass (0.75 cm thick) placed over the square hole forms the support surface for the gel cushion. The dimensions of the gel cushion are 40 cm x 40 cm x 3 cm. The X-Y positioning of the ultrasonic transducer is accomplished by two precision sliding assemblies driven by stepper motors under the control of an IBM CS/9000 computer. The positioning system is mounted underneath the seating surface.

In order to achieve good coupling and friction-free sliding between the ultrasonic transducer and the thin plexiglass plate, a special enclosed water coupling system was designed and fabricated. The coupling system includes a cylindrical cup for housing the transducer (Figure 1). An O-ring attached to the top surface of the cup facilitates smooth motion of the cup across the bottom surface of the thin plexiglass plate. A water tight seal is maintained by a spring loaded mechanism which gently presses the cup against the plexiglass plate. The cup is filled with water and connects to a small reservoir under pressure.

In order to obtain a three-dimensional contour, the ultrasonic transducer must be positioned at several discrete locations underneath the cushion. This is accomplished by moving the transducer along multiple scan lines in a rastor fashion. The three-dimensional system has been designed to accommodate a larger scanning area - to enable the imaging of regions around the trochanters as well as the ischial tuberosities.

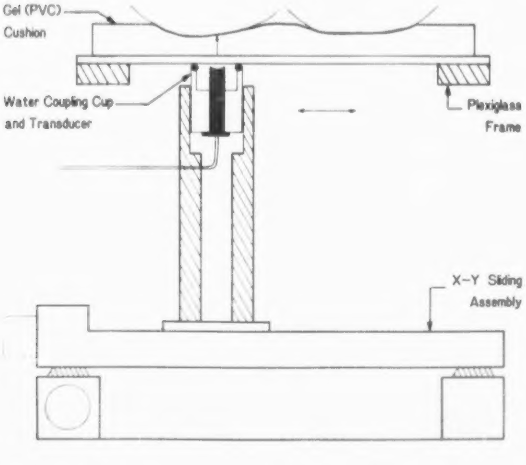

Figure 1. Schematic of the Scanning System

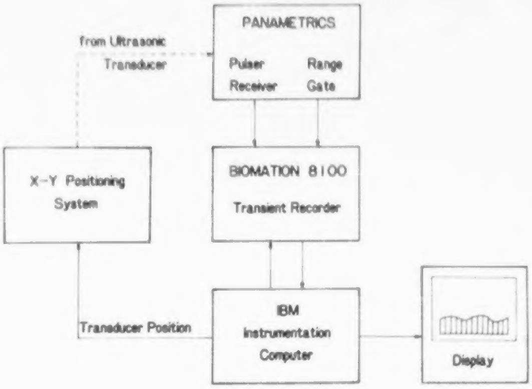

Figure 2. Ultrasonic Pulse-Echo System

The ultrasonic dimension gauging system (Figure 2) consists of a high energy pulser, receiver, calibrated attenuator, and a range gate housed in a single unit (Panametrics 5055UA). The electronic range gate facilitates the selection of a specific interface echo. A high speed analog to digital converter (Biomation 8100) is used to measure the temporal location of the interface echo. The excitation pulse along with the interface echo are sampled under the control of the IBM CS/9000 and stored on disks for further analysis. The distance between the excitation pulse and the interface echo is computed using a simple edge detection algorithm. The time delay is first computed in terms of number of sample intervals and then multiplied by the velocity of sound in the gel material (1400m/s) to obtain thickness. As the transducer moves along a scan line, each new contour point is displayed until the transducer reaches the end of that line. The resulting contour "slice" is displayed for examination before moving to the next scan line (Figure 3). After all the scans are completed, an overall contour is displayed using three-dimensional graphics on the IBM CS/9000. Using a 2MHz transducer, the resolution of the system in detecting interface position is approximately 0.70 mm.

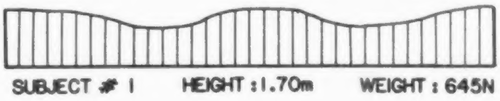

Figure 3. Simulated Contour "Slice"

In conjunction with the buttock-cushion contour information, pressure measurements are accomplished using a Pressure Evaluation Pad (PEP). The PEP is an electromechanical device comprised of a pad that contains a 12 x 12 matrix of pneumatically controlled flat contact switches placed 30 mm apart. The switches are encased in a Mylar bag and are interfaced to the IBM CS/9000 via a digital input channel. Switch closures are read by the computer and are displayed as a matrix

COMPARATIVE EVALUATION OF PRESSURE TRANSDUCERS FOR SEATING

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ABSTRACT

Three seating interface pressure evaluating systems have been compared for their measurement and performance characteristics. The pressure gauge for Scimedics, Inc., the pressure evaluating pad from the Texas Institute of Rehabilitation Research and the Oxford pressure monitor from Talley Medical Equipment Ltd., were calibrated using spot, plane and curved surface loading to simulate seating loads. Applied peak pressures could not be accurately reproduced by the transducers. Plane surface loading resulted in good agreement among the instruments but the other load types produced disagreeing results.

INTRODUCTION

Pressure distribution at the weight bearing interface is an important determinant of body support in wheelchair sitting. Devices used to measure pressure distribution ranged from a single transducer moved to new locations for successive measurements to whole arrays of transducers placed on the support interface for simultaneous observation of pressures. The pressure gauge from Scimedics, Inc.(1) represents the former while the pressure evaluating pad (PEP) designed at the Texas Institute of Rehabilitation Research (TIRR) (2) denotes a latter type. These systems employ electropneumatic techniques using inflatable vinyl bag transducers with internal electrical contacts for the detection of external pressure by internally applied equilibrium. Recently a simpler new system, the Oxford Pressure Monitor, (3) has become available from Talley Medical Equipment, Ltd. of England. This system operates on a completely pneumatic principle (4) which eliminates the possible breakage of electric conductors in the transducer bag.

The purpose of this work has been to compare the measuring and performance characteristics of the Scimedics pressure gauge, the TIRR-PEP which was parallel interfaced with the LSI-11/23 computer and the Oxford Pressure Monitor. Reproducible spot, plane and curved surface calibration loads were applied to each of the transducers to simulate some of the buttock-cushion interface loads. The gauge output was then recorded for several runs in each calibration for statistical analysis of the results.

METHODS

Three methods of loading were designed to investigate the transducer characteristics. The three transducers were loaded identically by each method. First, the load was applied in a spot to the transducers on a hard surface. Next, the load was applied by a plane and in the third the load was exerted by a curved surface. Spot loading was by a rigid 7.3 kg mass and 21.5 cm diameter bowling ball placed on the transducer on a plywood surface as shown in Figure 1. The contact area was measured separately by thin, sensitive tracer paper to be 1.2 cm diameter.

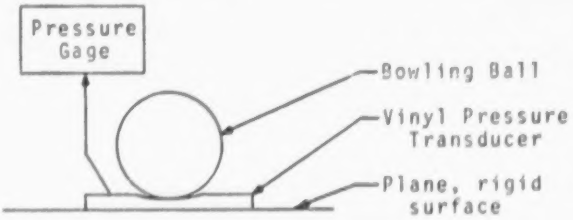

Figure 1. Spot load calibration.

Plane loading was between two parallel interconnected flat rubber envelopes which could be inflated to increasing pressures simultaneously. The loading area was measured to be 10 X 20 cm between the envelopes. The transducer was sandwiched between the envelopes as shown in Figure 2. Inflation of the envelopes applied the calibrating pressure to the transducer between the wood support plates holding the assembly.

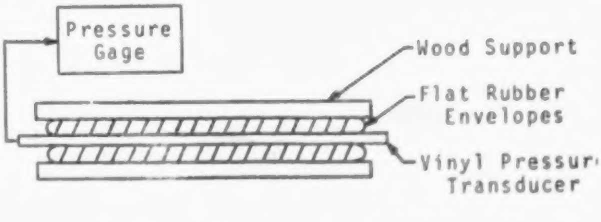

Figure 2. Plane load calibration.

The method for the curved surface loading is shown in Figure 3. The same bowling ball was placed on the transducer on a 3 inch thick polyurethane foam cushion. The cushion was indented by the ball 1.5 cm deep to make a contact area of 10 cm in diameter.

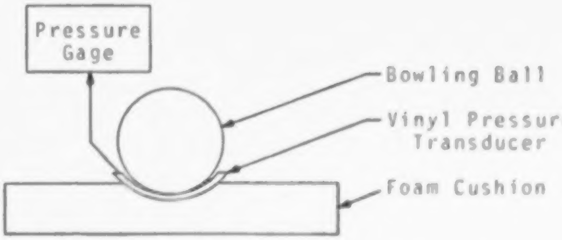

Figure 3. Curved Surface load calibration.

In each loading method the vinyl transducer bag was inflated until equilibrium was indicated between the external load and the pressure in the transducer. The transducer equilibrium pressure was then read with a standardized clinical sphygmomanometer at the resolution of ± 1 mm Hg. The circuit calibration of the Oxford Pressure Monitor was carried out according to the manufacturer's recommendation in a similar manner to the above mentioned spot calibration technique.

RESULTS

The measured pressure data from the spot loading calibration is shown in Table 1. This method of load application simulated the peak pressures applied through small contact area. The contact pressure was also derived by calculation using the mass of the ball and the measured contact surface between the ball and the transducer bag. The theoretical maximum contact stress between the ball and the hard surface was also calculated by the Hertz contact stress formula. (5) This calculation produced the possible contact pressure without the load distributing effect of the transducer bag. The ratio of the derived values in Table 1 shows that the transducer itself will reduce the peak contact pressure by nearly an order of magnitude. Comparison of the measured values shows the highest peak pressure attenuation by the Oxford Pressure Monitor.

TABLE 1 Spot load calibration

Maximum Peak Pressure (mmHg)

	Derived	Measured		
		TIRR PEP	Scimedics	Oxford
Hertz contact stress	$4.7 \cdot 10^4$			
Load per area	$6.8 \cdot 10^3$	200	138	68

All transducers exhibited position sensitivity. The location of the contact spot relative to the center of the sensor was most critical with the TIRR-PEP and least with the Scimedics pressure gauge, probably reflecting the relative size of the individual sensitive areas of each transducer.

The pressure values developed on the indented curved surface of the foam cushion by the rigid bowling ball are shown in Table 2. The Hertz contact stress was again derived by omitting the effect of the transducer from the calculation and also the contact pressure was calculated by the load per area method. The nearly 50% increase in calculated pressure indicates that the vinyl transducer bag stiffens the interface between the foam and the ball and results in elevated pressure. The percent change between measured and derived values probably represents the material and structural differences between the vinyl transducer bags.

TABLE 2 Curved Surface Load Calibration

Mean Contact Pressure (mmHg)

	Derived	Measured		
		TIRR-PEP	Scimedics	Oxford
Hertz contact stress	45.8			
Load per area	68.3	54.1±1.0	73.3±1.1	92.3±0.1
% Change	49.1	20.8	7.3	35.1

Calibration by the plane surface loading resulted in the best agreement between the externally applied load and the measured values as shown in Table 3. The results also indicate the high degree of reproducibility that each transducer is capable of with uniformly applied plane loading.

TABLE 3 Plane Load Calibration

Applied Pressure (mmHg)	Measured Pressure (mmHg)		
	TIRR-PEP	Scimedics	Oxford
20	20±1	20	15.6±1.0
40	40±1	40	39.2±1.0
60	60±1	60	60.8±1.1
80	80±1	80	81.1±1.8
100	100±1	101	102 ±1.4
120	120±1	122	122.3±1.0
140	140±1	141	141.4±1.8
160	160±1	162	159.5±1.7

CONCLUSIONS

The application of pressure transducers was found to be valuable for laboratory and clinical determination of pressure distributions at the wheelchair cushion interface. None of the transducers was found to reproduce the applied peak pressures in the calibrating modes of this experiment. The difference between the measured and calculated peak loads could be ten fold. The shape of the load was found to be critical for the accuracy and the reproducibility of the sensors. The measurement time for a survey of pressure distribution on a wheelchair cushion could be a lengthy process. Using the Scimedics pressure gauge for pressure measurement at 7 different cushion locations required 5 minutes at ± 1 mm Hg resolution while the other two transducers at the same resolution needed 20 minutes or more.

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ACKNOWLEDGMENT

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20.6

A PRELIMINARY ANALYSIS OF LIMB GEOMETRY AND EMG ACTIVITY FOR FIVE LEVER PLACEMENTS

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ABSTRACT

Changes in EMG activity of selected muscles corresponding to limb segment (LS) positions for five seat to lever placements were analyzed for simulated wheelchair (WC) propulsion. The various seat positions resulted in large differences in LS amplitudes and positions relative to their anatomical axes. EMG activity varied in a similar manner. It was found that "extreme" positions required greater muscle involvement, particularly in the recovery phase of the stroke cycle (SC).

INTRODUCTION

It has been demonstrated that seat to propulsion interface can have a significant influence on WC propulsion efficiency.(1,2) While this effect was shown to be more pronounced for rim propulsion the range of positions for lever propulsion was not large.(2) The selection of lever propulsion for this study was based in part on the potential of this method of WC propulsion, but more importantly, due to its greater consistency and mechanical simplicity. The physical propulsion components, i.e., the arm, forearm and lever, may be viewed as a 4-bar linkage with the fixed positions of the shoulder and lever axes (LA) acting as the 4th link. The complexity of this mechanism is increased by the extra degrees of freedom at the shoulder and wrist to lever articulations. The optimization of this "4-bar" system depends upon the relationship of LS geometry to muscle function. It is the purpose of this study to explore this relationship and induce the probable effects of subject factors relative to position.

METHODS

An able-bodied 18-year old male, 167cm. tall and weighing 58kg. served as the subject of this study. The selection of this subject was based on his relatively small stature, which magnified the position variation, and his sensitivity to Faradic muscle stimulation. The subject was exceptionally lean and muscular which facilitated muscle identification and isolation. Eleven muscles (see Fig. 2.) were monitored during simulated WC propulsion at the nominal speed and power of 3.89km/hr and 15W, respectively, for each of 5 positions. The positions were determined with respect to the horizontal and vertical distances from the LA to the rear seat post intersection. Horizontal seat variations included 0, 6 and 12in. aft of the LA at a vertical height of 6in. This corresponded to respective shoulder to LA dimensions of -7.7, 7.5 and 22.7cm at a vertical distance of 69.2cm. The vertical variations were 2, 6 and 10in corresponding to shoulder to LA dimensions of 59, 69.2 and 80cm, repectively, at the middle horizontal position. Motion of the LSs was determined with stereometric position sensors (PS). The analog data outputs for EMG and from the Dynamometer and PSs were sampled though an A/D converter and stored on diskettes.

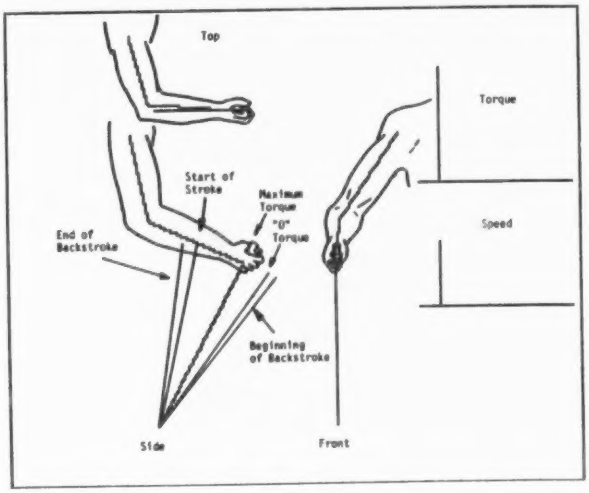

Figure 1. LS Positions with Corresponding Torque and Speed.

RESULTS

The positions of the LSs at the start of the stoke cycle (SSC), at max. torque (MT), end of torque cycle (ETC), lever reversal (LR), forward lever reversal (FLR) and end of cycle (EC) were graphically reconstructed for each of the 5 placements. These illustrations are shown in Fig. 2. The various positions and views are identified in Fig. 1. The average EMG activity corresponding to these various phases of the SC and at intermediate deciles during the recovery phase (RP) are also shown in Fig. 2. The respective percentages of the SCs for the above positions and stroke periods are shown in Table 1.

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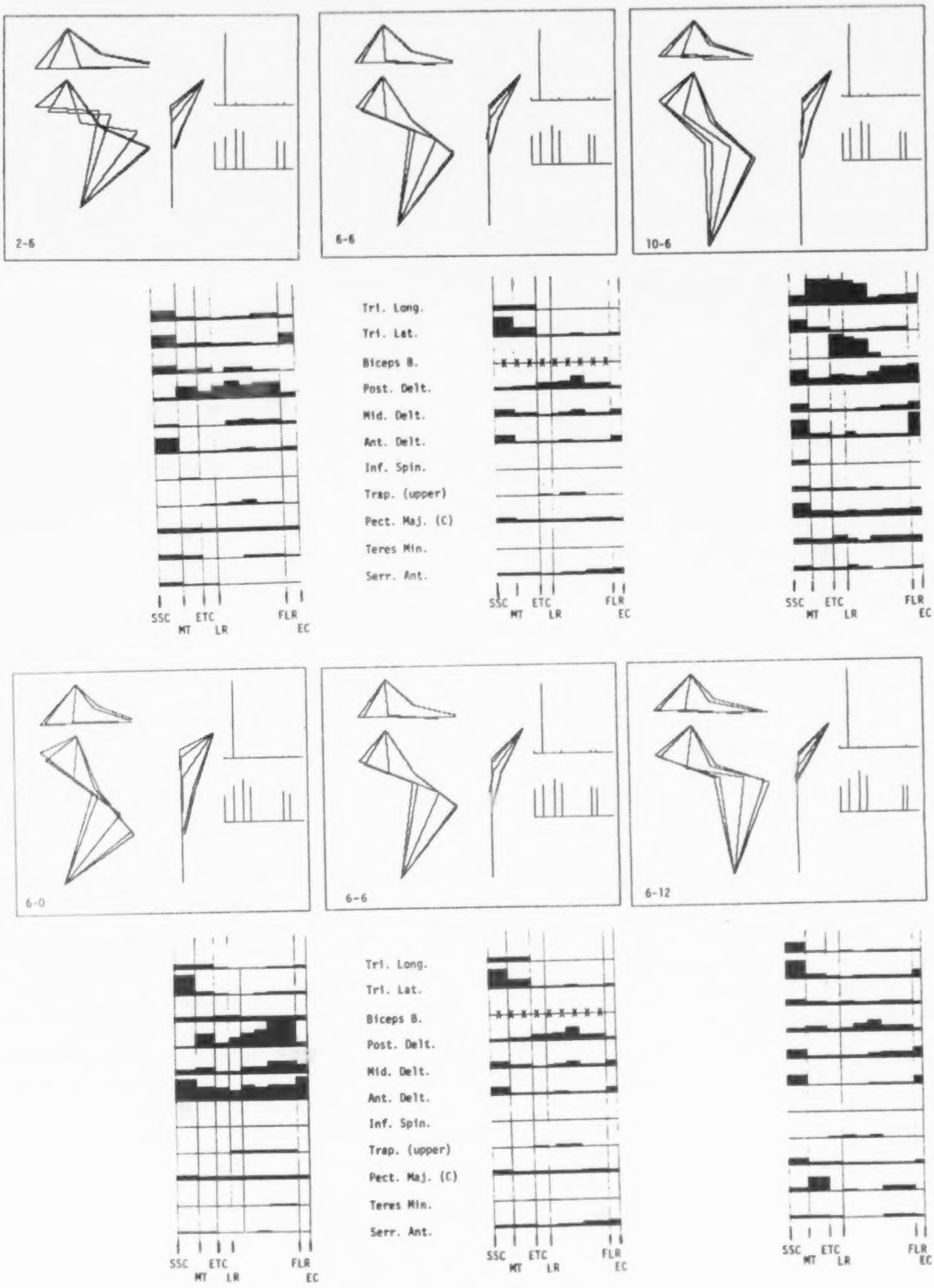

Figure 2. Comparisons of LS Positions and EMG Activity for 5 Seat Placements.

PLACEMENT	PERCENTAGE OF SC						PERIOD
	SSC	MT	ETC	LR	FLR	EC	(SEC.)
6-12	0	14	30	41	94	100	.91
6-0	0	16	30	41	91	100	.86
6-6	0	15	33	45	92	100	.92
10-6	0	13	31	43	92	100	.88
2-6	0	16	31	42	90	100	.96

Table 1. Percentages of Stroke Cycle.

An inspection of Fig. 2. reveals that overall EMG activity is at a minimum for the 6-6 placement followed by the 6-12 placement. This is also respectively the case for the EMG activity during recovery phase (RP) of the SC. The EMG activity is particulary high during the RP for the 6-0 and the 10-6 placements. Although the stroke amplitudes for the 2-6 and 10-6 appear to be the largest and smallest, respectively, this quantity seems remarkably constant over the 5 placements. It cannot be determined from Fig. 2., but both the 2-6 and 10-6 placements result in excessive internal rotation of the arm and the 2-6 in a very acute forearm flexion angle. It can be seen from the Front views in Fig. 2. that the 6-0 placement results in greater arm abduction than any of the other placements. The EMG activity appears to confirm that the Ant. Delt., Pect. Maj. and the Triceps, particularly the lateral head, are the principle driving muscles for lever propulsion.

DISCUSSION

The absense of data from other subjects dictates a conservative approach in interpreting the results of this study. Further caution is warranted by the limitations imposed by the use of surface electrodes, particularly so for the Inf. Spin., Teres Min. and Serr. Ant. Despite these limitations the results based on EMG activities of the Deltoid and Triceps and the lower overall activity for the 6-6 and 6-12 positions are consistent with A Priori knowledge of the relationship of Propulsion Efficiency to Placement. These results are further consistent with the hypothesis that a significant part of the Efficiency loss in WC propulsion occurs in the recovery cycle. The excessive angles noted earlier for arm abduction, internal rotation and forearm flexion are also consistent with reduced efficiency as a result of having to perform work with the muscles shortened. These results are encouraging. While the task of determining the effect of various variables on WC propulsion Performance continues to be a formidable one, it is evident that a logical determination of the effects of seat placement on muscle function, and consequently on performance is possible.

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A MICROCOMPUTER BASED MYOELECTRIC LIMB CONTROLLER

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ABSTRACT

This paper describes the design, implementation, and testing of a microcomputer based myoelectric limb controller. The resulting controller was constructed for use as a development tool to help explore clinical microcomputer based myoelectric limb controller applications. The present device uses low power CMOS technology to provide a system in which most of the new computationally and/or decision orientated myoelectric signal processing algorithms can be used. Two CMOS microprocessors, a CD80C86 and a MC146805E2, are used but the total limb controller power consumption is only 410 mA. The controller can provide many advantages over present analog myoelectric limb controllers.

BACKGROUND

While many of the advances in powered upper limb prosthetic limbs have increased either the availability, functionality, or cosmetic appeal of prosthetic limbs they all rely on essentially the same method of two state control that was developed in the late 1940's and early 1950's [1]. Some of the present day myoelectric limb controllers use three state control which was first developed in the late 1960's [2]. Since then some five state control schemes have been proposed [3], however, no clinically useable systems have resulted. Therefore, the use of multilevel muscle control appears to have reached an impass. In response to this problem new control algorithms are under development, but generally these are computationally and/or decision orientated. Therefore, they cannot be used in present myoelectric limb controllers because present limb controllers use analog circuits to operate. Thus, a myoelectric prosthesis controller is needed on which these various control schemes can be easily implemented. A computer based limb controller would allow this flexibility. Here, complex and simple control algorithms could be implemented by merely changing the limb controller software. The design, implementation, testing, and demonstration of such a myoelectric limb controller using microprocessor based computer hardware was the objective of a project [4] recently undertaken at the Rehabilitation Center for Children and the University of Manitoba located in Winnipeg, Canada.

LIMB CONTROLLER OVERVIEW

The goal of this project was to create the necessary hardware and software to implement a variety of modern myoelectric prosthesis control algorithms in order to investigate the feasibilty of myoelectric prosthesis control using microcomputers. However, it should be noted that the limb controller is intended as a demonstration project and design tool and not as a final end product. A block diagram of the limb controller setup is given in Figure 1. Figure 2 shows a photograph of the limb controller with the prosthetic limb.

Presently the limb controller consists of five printed circuit boards each approximately 23 cm. (9 in.) square. These dimensions preclude its immediate use in a clinical situation, but the limb controller described here is a developmental system from which other microcomputer based limb controllers will evolve. The overall size was purposely exaggerated to allow easy testing and modification of the limb controller. Using modern circuit miniaturization techniques, such as very large scale integration (VLSI) and component hybridization, the dimensions could be reduced to a clinically usable package. The device is powered by a 7.2 volt battery with current drain of about 410 mA.

Four channels of myoelectric signals can be acquired at a maximum sampling rate ranging from 10 kHz for one channel to 2.5 kHz for all four channels. The limb controller microcomputer is made from two microprocessors; one, a CD80C86, is dedicated to signal analysis, while the second, a MC146805E2, controls the data acquisition of the myoelectric signal and the prosthesis motors. Motor control is presently only "open/close" in nature but modifications to the limb controller software can provide proportional control. Limb position feedback has also been incorporated into the limb controller with ongoing research into effective low power transducers for joint positions. The above hardware allows the limb controller to operate in real time using a variety of control algorithms.

TESTING: RESULTS AND DISCUSSION

Test programs on the limb controller microcomputer insured that the hardware operated in a correct manner. However, the limb controller software demanded more extensive testing. Two and three state control was tested by first verifying that the control algorithms could be correctly operated by the authors. After this several subjects were used to verify that the device could be operated by a range of users.

The advantages of a microcomputer based limb controller become apparent when the abilities of this controller are compared to those of conventional myoelectric limb controllers. The two and three state control algorithms were programmed and tested in just several days compared to the several weeks that it would take to develop this control in a dedicated hardware form. One of the real strengths of microcomputer control lies in the "intelligence" (i.e. flexibility, complexity, functionality, etc...) that can be programmed into the limb controller. This "intelligence" ranges from the ability to detect spastic contractions to turning off the joint motors when a joint range limit has been reached. Several preprogrammed multiple joint movements were programmed and demonstrated to show the increased prosthesis functionality that can be added with microcomputer control. However, the limb controller's strongest advantage lies in its ability to implement and test the new myoelectric control algorithms.

SUMMARY AND CONCLUSIONS

The implemented limb controller described here demonstrates that a microcomputer based myoelectric limb controller is feasible. The limb controller provides many facilities that conventional myoelectric limb controllers do not. Once the hardware on this controller was debugged and operational the present technology for limb controllers (i.e. two and three state control) was implemented in software in approximately three days. As well, this could be customized for each user with quick and relatively simple software changes. Features such as preprogrammed multifunctional movements, superimposed movements, and limb position feedback cannot be duplicated by any other non-computer based limb controllers.

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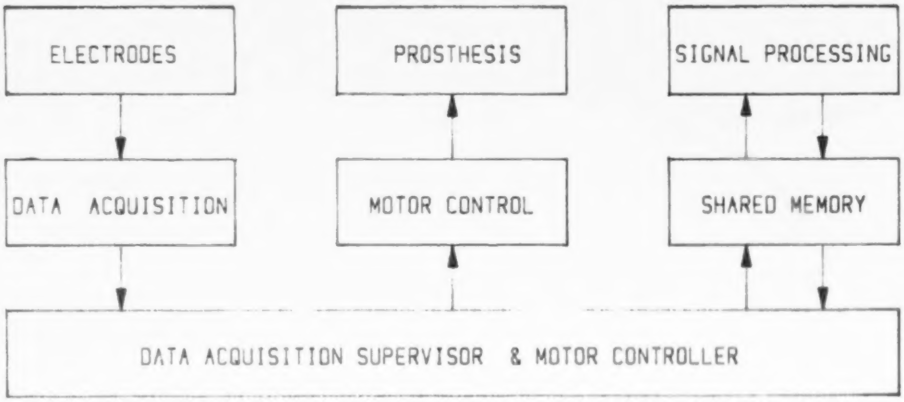
ELECTRODES
PROSTHESIS
SIGNAL PROCESSING
DATA ACQUISITION
MOTOR CONTROL
SHARED MEMORY
DATA ACQUISITION SUPERVISOR & MOTOR CONTROLLER

Figure 1. Block Diagram of Limb Controller.

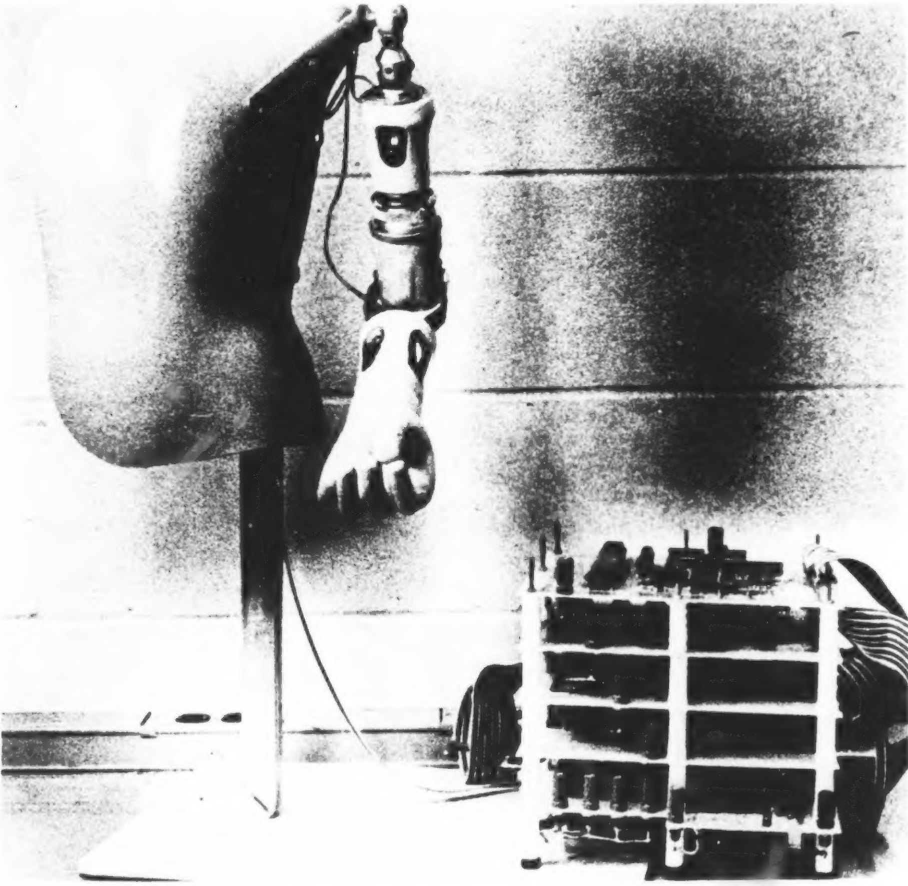

Figure 2. Limb controller with the Prosthetic Limb.

21.2

A DATA-LOGGING SYSTEM FOR WHEELCHAIR USE SURVEYS

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ABSTRACT

This paper describes a data-logging system for use in wheelchair surveys. The data collection system is currently configured to record the forward and lateral accelerations, time used, total distance travelled, average and maximum speed, and the number of uses during a one-week test.

The system is built around a battery powered recorder unit that is mailed to the remote wheelchair site, where a technician mounts it onto the wheelchair under test. After collecting data, the recorder is returned, and the results are stored in a computer for later analysis.

Devices that only record distance and time of use cost over $200. The system described does more for less than one half the price per recorder.

INTRODUCTION

The system shown in Fig. 1 is composed of three parts (1). The portable recorder in Fig. 2 mounts onto the wheelchair and collects readings from its sensors. When the test is over, the recorder transfers its results to the host computer, where it is stored on floppy disk and analyzed. The recorder connects to the host through an interface unit that synchronizes the two, and performs the voltage-level translation needed to allow them to communicate.

To use the system, the technician initializes the portable recorder, seals it, then mails it to the remote wheelchair site. The technician at the remote site mounts the recorder onto the wheelchair. The recorder automatically monitors the wheelchair when the user sits. The recorder also shuts itself off when the user leaves. After the user has used the wheelchair for about a week, the technician returns the recorder. The information is then loaded from the recorder into the host computer, where it is later analyzed.

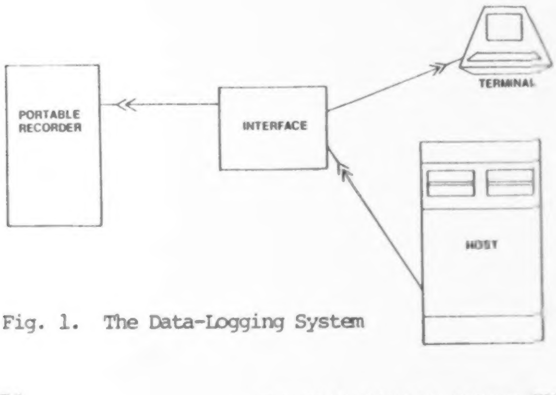

Fig. 1. The Data-Logging System

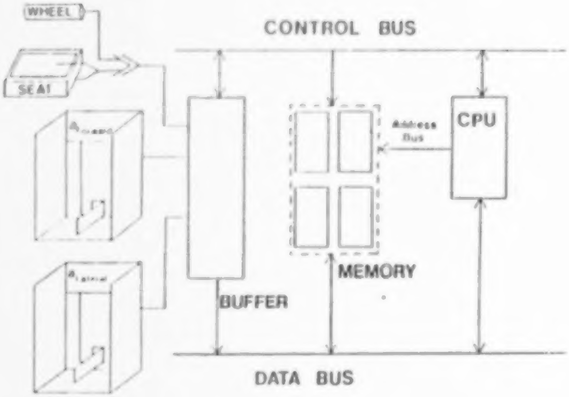

Fig. 2. The Recorder

Figure 3 shows the results of a sample test.

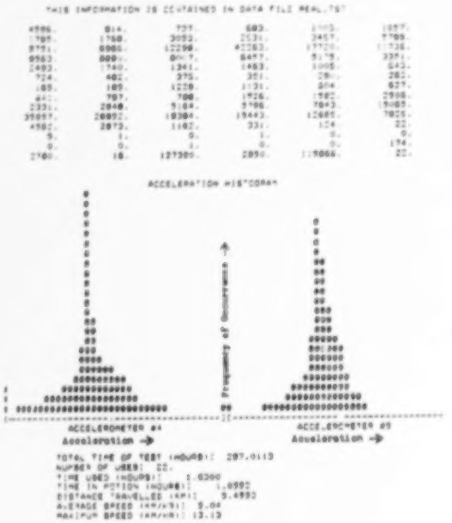

Fig. 3. Output of a Sample Test.
Shown are the data stored in the file, the acceleration histograms, and some usage parameters.

THE RECORDER

The portable recorder that mounts on the wheelchair consists of a microcomputer, sensors (described later), and interface circuitry, which has five analog inputs. The entire device consumes about one milliamp from a set of four "C" cells.

All recorder functions are controlled by a 512-byte microcomputer built around a CDP1802 8-bit CMOS CPU (2). The computer has two interesting features. The first is that the entire program is stored in volatile RAM. This allows the recorder to easily adapt to new test conditions. To reconfigure the recorder, the technician simply adds the new sensors, and modifies the program. (The recorder's programs are stored in the host computer and downloaded to the recorder before a test.) The other interesting feature is the computer's variable-rate system clock. The slow speed lets the computer consume less power during idle periods (which is most of the time), and it is also used when communicating with the host (because of the signal timing considerations). The computer selects the faster clock rate so that it may handle the greater processing burden associated with collecting data.

THE SENSORS

The recorder currently monitors four inexpensive, custom-designed sensors. They are: the seat sensor, the wheel sensor, and a pair of inertial accelerometers. The seat sensor is a flat, pressure-sensitive resistor that is made out of conductive foam. The computer uses it to determine if the wheelchair is in use. The wheel sensor is a magnetic door switch, usually found in burglar alarms, that opens when a magnet mounted on the wheel passes by. The computer uses the wheel sensor for the purpose of counting wheel revolutions.

Figure 4 is a drawing of one of the flexure (spring-mass) accelerometers used in the recorder. The mask partially blocks light from an IR-LED to a phototransistor. The computer calculates the acceleration based on the output of the phototransistor. It has a range of ±1 g. This sensor design was chosen because it is more stable than a piezoelectric accelerometer, it uses less current than a closed-loop servo one, and it can be inexpensively built.

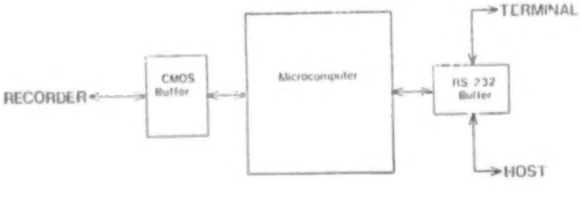

Fig. 4. The Interface Unit.

THE HOST

The host used is an RD-11 computer, which is compatible with the PDP-11. Software was written for the purpose of initializing the recorder, storing test results, and printing test data. A complete program development system was written for creating programs used in the recorder. All host software was written in FORTRAN IV, so it is transportable.

THE INTERFACE UNIT

The recorder connects to the host through an interface unit shown in Figs. 1 and 5. The interface unit connects to the recorder, and between the terminal and the host's terminal port. Its function is to provide the handshaking and the voltage conversion needed for communication.

It works by monitoring the serial bit stream from the host and terminal. When it sees the proper characters, it then controls the information flow between the recorder and the host. Communication with the host is by means of the RS-232 protocol. DMA is used to access the recorder's memory.

The interface is built around an Intel 8748 single-chip microcomputer (3) and buffer circuitry. It has a selectable baud rate, and it can drive a terminal when the host is not needed to store the data.

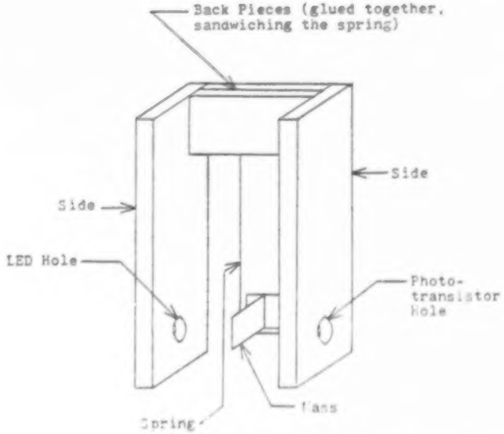

Fig. 5. One of the Inertial Accelerometers.

EXPERIMENTAL RESULTS

The accuracy of the time, speed, and distance measurements was verified using treadmill testing. There was no means of absolutely confirming the acceleration results, but they appeared to be reasonable. Carbon-zinc batteries supply plenty of power for the duration of an entire test (about one month, including postal transit time).

CONCLUSION

The data-logging system is both a powerful and inexpensive alternative to the currently existing data loggers on the market. The system requires no special hardware to connect to any computer with a serial terminal port. The analog input channels and the development system allow the recorder to be easily modified by the user for customizing it for different surveys. One researcher already plans to add temperature sensing to monitor seat comfort.

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COGORTH: A PROGRAMMING LANGUAGE FOR COMPUTERIZED COGNITION ORTHOSES

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INTRODUCTION

Patients who acquire diffuse and/or focal lesions of the brain often sustain dramatic and potentially debilitating changes of cognitive functioning. These changes may be characterized by limitations of attention, orientation, memory functioning, reasoning, social skills, and higher order integrative functions. They may lead to severe disruptions of behavioral style, level of independence, interpersonal relationships, and vocational capabilities. A technique for assisting such patients to function independently using a computerized "cognition orthosis" has been developed (1). It has been successfully demonstrated to guide brain injured patients through tasks which they could not otherwise perform unaided.

This paper describes a) the basic features of the programming language, COGORTH, specifically created for the development of computerized cognition orthoses; and b) a conceptual framework for developing COGORTH programs (Instructional Modules).

THE COGORTH LANGUAGE AND INTERPRETER

COGORTH is a specialized computer language providing a highly structured environment for programming sequential messages. These messages can be used to assist patients who need guidance for the completion of complex activities. They can be in the form of text presented on a video display, an audio signal, or a visual cue such as a flashing light. COGORTH's branching capabilities permit transition between different sets of messages.

A COGORTH program can display directions to a patient at any level of specificity, for any task which can be reasonably represented as a sequence of steps. It can present sequences of messages at any time of day specified by the programmer, and can repeat this sequence of messages at any interval. For instance, a series of messages instructing a patient to take a medicine can be presented any number of times a day, everyday, for as long as the programmer desires.

It is envisioned that COGORTH will be used by health professionals having a wide range of programming skills. Careful consideration has therefore been given to balance the power and complexity of the language vs. the need for simplicity. This has led to a design approach whereby each function or command can be: a) simply employed using default parameter values invisible to the programmer or b) used in a more sophisticated manner allowing detailed control of command or function implementation.

COGORTH provides programming capabilities for Instructional Modules (IM's) which can:

a) check a patient's performance for errors;
b) branch to error correcting or help procedures to assist patients when difficulties or errors are encountered;
c) manage interruptions of a task when a higher priority task must be completed or the allotted time has expired;
d) manage a patient's environment through control of electric applliances, telephone, and audio signals.

Error Detection

A message or request for keyboard input from a patient can be used to obtain feedback regarding their performance. A great deal of error checking can be performed on the basis of a patient's response (i.e. a patient can be asked to count the number of pills remaining in a bottle to check whether a pill has been taken). An inappropriate response can signify difficulty and the need for extensive help. A lack of a response within an allotted time period may also indicate a problem.

Branching.

COGORTH permits branching from any one message to another. In its simplest form, a message sequence can proceed in a linear fashion until a certain task is completed. In more complex sequences, intricate branching may be required to correct patient errors, to limit the amount of time permitted for a particular stage of a task, or to interrupt one task by another of higher priority. The branching feature of COGORTH is crucial to the development of even the simplest IM. One of its most common uses is in developing help and error-correction routines:

Error correction provides a patient with feedback about unacceptable responses. For example, an IM may call for a "yes" or "no" response. If a patient responds incorrectly, the IM might branch to a generalized routine which indicates to the patient that an error has occurred and assists them to respond correctly. In other situations an IM may branch to an extended help routine and then back to the main process if and when appropriate.

Interruptions.

COGORTH permits the programmer to design IM's so that one sequence of messages can be interrupted by another, if the interrupting sequence has been assigned a higher priority. This feature is particularly important since patients may be required to perform tasks which have some urgency (answering the telephone, removing boiling water from the stove, taking medication, etc.). Such tasks must typically be completed before continuing with the interrupted task. Two types of interruptions, absolute and relative, are available. Absolute interruptions occur at a pre-determined time of day. Relative interruptions are specified as an interval of time relative to some action or message presentation. For example, an absolute interruption would be useful in programming a message for taking medication at regular times during the day. A relative interruption could be used to signal a patient to remove an item from the oven.

Environmental Control.

Environmental control features include audio signals and (BSR) control of electrical devices (i.e. lights) which can be used to call a patient to the computer. The control of electrical devices also permits certain safety features (e.g., turning off an iron) should a patient fail to respond to a request from the computer. Telephone communications control allows the monitoring of incoming calls (to prevent unwanted interruptions) as well as notification of outside individuals should a problem condition be identified.

INSTRUCTIONAL MODULE DESIGN: CONCEPTUAL FRAMEWORK

Instructional Modules (IM's) are COGORTH programs which guide patients through various activities. They contain lists of visual and auditory messages which are presented in sequential order. IM's must be constructed using messages that are consistent with the patient's level of cognitive functioning, recognize errors, provide for error correction, and act upon errors which are not correctable (i.e. reqest outside intervention).

A basic design structure for IM's is important because even a simple activity may require a relatively complex IM to recognize and perhaps correct for a user's response. A conceptual framework has therefore been adopted (2) for systematic representation of complex activities in IM's.

Briefly, complex activities are treated as behavioral units called processes. A process is any reproducible sequence of activities which can be systematically and comprehensively reconstructed as a series of stages. Each stage in turn can be represented as a set of behavioral options from which an individual chooses. (A formally complete description of a process includes options which account for failure as well as success). The particular pattern of options correctly or incorrectly chosen by the individual is called a version of the process.

For example, baking a cake with icing can be conceptualized as a task having two processes, baking the cake and preparing the icing. Each step, such as batter mixing or removing the cake from the oven, can be considered as a stage. Within each stage are a number of options (such as the choice of mixing the batter with a spoon or food processor). Some options can actually change the order of stage completion (i.e. preparing the icing while the cake is in the oven vs. waiting until it is done). The application of the conceptual framework described above to a task assists in the development of the IM by providing a standardized structure. It also makes it feasible to employ generalized help and error correcting routines based on the conceptual framework and its imposed structure.

DISCUSSION

Successful compensatory techniques which use environmental cuing such as a cognition orthosis can increase a patient's level of independence, decrease the amount of supervision and guidance required, and potentially improve quality of life (3). A patient's mood and self esteem may also be enhanced by compensatory techniques which broaden the range of functional activities in which the patient can effectively participate.

The COGORTH language provides the specific capabilities required to construct an effective cognition orthosis. In essence it is an authoring system for instructional modules to be used in guiding patients through a variety of tasks. The necessity for a custom programming language arises from the variety of tasks and wide range of patients for which IM's are to be developed. The COGORTH language has been designed for use by health professionals with little programming expertise yet provides powerful options for more experienced programmers.

Current work is now focusing on the development of IM's for vocational applications and the investigation of remedial effects from the use of a cognition orthosis. A library of IM's for a variety of tasks is also planned. These modules will be able to present messages and guidance at different specificity levels for adaptation to the needs of various users. Thus, while some custom IM development will be necessary (by health professionals with the necessary expertise), library modules will be available to minimize individual programming and customization.

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HEALTH AND HUMAN SERVICE ROLES FOR ROBOTS

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ABSTRACT

Evolving robot technology can play a pivotal role in helping to maintain humans in a more independent role in both home and institutional health care delivery settings. Although the useful robot —from a health and human services perspective— has not yet evolved, robotic technologies are rapidly moving toward utilitarian applications in this important area. This paper will present an overview of some of the issues that will need to be addressed as robots evolve into service roles for humans.

BACKGROUND

Robots

Industrial robots and personal robots are poles apart today. However, this schism cannot continue,[5] stated Joseph Engelberger in a recent editorial. The concept of robots has been popularized since Karel Capek wrote the play R.U.R. in 1920. Industrial robots first appeared on the market in 1959 and have continued to flourish since. *Personal* robots became widely available on the market in 1983 and, in April, 1984, the First International Personal Robot Congress & Exposition was held in Albuquerque, New Mexico.

Even though some people may have difficulty imagining the *hulking metal creature bolted to the factory floors* and the friendly R2D2 or snowman shapes emanating from the emerging personal robot industry as having much in common, our work over the last half decade has demonstrated both conceptually and practically the importance of combining certain characteristics of both types of robots. The *interactive robot* represents a nascent move toward the development of a utilitarian product to serve humans in a wide range of applications.

From the health and human service perspective, one of the most important issue for the next decade will be the meshing of the industrial and personal robot markets. *Industrial robots are gaining sensory perception, advanced language and mobility; and so are the hobby robots. With so many bright enthusiasts in the game the two robotics cultures must converge. Before this decade is out it should become evident that robots will no longer need the distinction* **industrial** *and* **personal**, *they will all be just robots "under the skin."*[5]

In the future, the rate of introduction of robots into the workplace will increase. Applications areas will grow, and many currently unthought of services will be provided via robotic technology and its spinoffs. There is probably no potential robot market larger or more challenging than the health care and human service industries. Since the advent of modern medicine, the medical community has been open to new ideas to help diagnose, treat and alleviate human fraility. Technological advances have the potential for significantly enhancing the ultimate rehabilitation goal of *self-sufficiency*. First, we must begin to define "human services" in the broadest perspective. The same robot created for in-home use to serve disabled individuals could also be the "household" robot that will be used to serve persons with a far greater range of abilities. A wide market for these household robots will help to bring the costs down as **economies of scale** come into play.

DISCUSSION

Who will use a health and human service robot?

The potential market is rich; in effect, it will encompass all **humans** who want, need, or require services. Specifically, in health services, candidate users will include allied health care professionals, maintenance staff, housekeeping staff, and patients— both children and adult. For instance, a nurse might use a robot to assist in a patient lift or transfer task. A physician might use one for monitoring patients or as an *expert* assistant. A patient might use a robot for fetching or carrying tasks. Housekeeping might use a robot to help with delivery and cleaning chores and physical therapists might use one to administer range-of-motion exercises.

Who will develop and provide health and human service robots?

In some industries most commercially successful products are developed by product users, not product manufacturers.[17] Industrial and personal robot **users, manufacturers, universities, government** research centers, as well as **hobbyists** will (in the future) be the major sources of robotic innovations for health and human service applications. Presently, industrial robot manufacturers are generally unaware of potential applications of their technologies in health and human service settings. Applications areas will need to be defined and this information transferred to interested manufacturers.

It is becoming increasingly important to develop joint research efforts which include potential **end-users** from the onset. Methodologies are being developed to help facilitate user involvement in the design/dissemination lifecycle of a robotic aid at the Palo Alto Veterans Administration Medical Center.[7][8]

What will these robots look like and **What** will they do?

> *Packaging for specific environments will require that the system be modular and flexible so that (a)it can be adapted to different needs in different installations, (b)it can be readily changed as new sensors or data management procedures become available and (c)it can be modified as new practices for treating patients are introduced. The modularity requirement pertains to both hardware and software.*[4]

First, we might think of the robots that exist at the present time as comprising a "gene pool" of **robot characteristics**. That is, a *pool* of technological characteristics from which designers and developers can draw. This pool might include characteristics as diverse as the dexterity and precision of the industrial robot to the *personality* of the personal, hobby, or toy robot. Cross-fertilization will result in robotic systems which are unique blends of existing components of hardware and software. Developments in the area of artificial intelligence, such as languages, knowledge programming, visual graphics, expert systems and next generation software will need to be closely monitored for advances that could be used for applications in health and human service areas. *The next several years will see an explosion of sensors and sensor applications.*[4] These advances will add a previously unavailable 'potential' to this robot gene pool. It is important to note that probably no other market offers a greater demand for the symbiosis of *high tech* and *high touch* than health and human services. We can begin to work toward developing 'hybrid vigor' which has consistently been a predictor of success in biological systems. The fusion of defined needs/desires with the appropriate application holds promise for spurring the creation of more versatile systems to serve human demands.

Second, we must consider the optimal **environment** in which these expressed (chosen) characteristics can be utilized. Robotic systems will evolve to fill *environmental niches*. Perhaps we can analogize robot evolution to species evolution. Different forms will evolve to fill niches best suited to their particular range of *skills*. No less than in the case of their living counterparts, robotic species differentiation and successful survival are a result of *taking advantage* of environmental niches. The more successful are the ones in *optimal settings* best suited to their individual strengths and minimizing their weaknesses. Characteristics that are useful in a particular set of environmental demands will be retained and refined over successive generations.

Robot evolution might also be compared with the evolution of transportation technology. We have foot powered, arm powered, animal powered, water powered, air powered, electric powered, cable powered, and gasoline powered vehicles, each one best adapted to suit our various mobility needs, desires, and capabilities. The important point to make is that, likewise, there is a whole spectrum of needs in the health and human services arena that can be met by innovative uses and applications of robotic technology. Robots that are developed with the most appropriate or *fittest* set of design characteristics will have an advantage for **survival** in the *competitive* market environment. It is our goal to identify those niches that need to be filled and determine what characteristics a robot system must have in order to optimize its performance in the specified setting. These niches will comprise a complex multi-dimensional set of requirements. Both machine and user characteristics, as well as task definitions, will need to be clearly analyzed for specific niches.[6][7][8]

> *Human factors can contribute to robotization in two areas: development of robot technology and allocation of functions among humans, robots, and machines.*[15]

Third, we must objectively consider and measure the *goodness of fit* of the robot to its projected niche. Design and human factors research must merge with marketing and manufacturing considerations to produce viable robotic systems. **Human factors** research can provide a critical feedback link to designers and developers of robotic technology. Assessment of how humans perform designated tasks and what portion of the tasks could be augmented are the first steps to assessing a potential fit.

For the most part, researchers, scientists, and engineers know little about the action sequences used in everyday tasks. *There is no set of mathematical equations to describe the kinematics of the human body, it is simply too complex.*[2] Add to this complication the perception of Koncelick that *there is no such thing as a normal adult or an average person,*[11] and one can begin to understand the enormous challenges that face designers and developers of systems that might replace, augment, or mimic human motions. End users can play pivotal roles in the design of these proposed systems. Consideration of robot factors in tandem with a wide range of human factors during the design, development and evaluation phases will yield a human-robot system that is more compatible with its environmental niche. This is not the path that has traditionally followed to produce an innovative product. Too often the product is designed and developed in isolation from the proposed end-user and/or defined applications.[17] Evaluation during developmental stages has also been omitted. The product reaches the market without being designed appropriately for the purchaser. The result is that after the novelty has worn off, usage rates inevitably fall to unacceptably low levels, or outright rejection of the device may ensue.

To use this evolving robotic technology effectively, there will be an increasing need to identify the critical points of impact. The identification of **tasks** that will need to be performed and the aspects of the task that will be augmented by the robot must be clearly defined and quantitatively described. Tasks consist of various sets of actions[9] which need to be replicated. We need to know answers to questions such as: how much force is required to perform specific tasks related to human functioning. For instance, questions such as: at maximum arm extension, for selected age groups and for selected robotic arms, what is the total payload required for Activities of Daily Living, vocational and recreational tasks? The next step is to apply these human factors findings to the design of robots.

The inclusion of relevant human factors input into the design and development of 'robot factors' can serve to strengthen the quality and versatility of the end product and make it a more attractive prospect and a more appropriate tool for a wider spectrum of end-users.

Mistakes are costly. Recent technology has produced new tools for design and manufacturing. Computer aided design and manufacturing (CAD/CAM) and simulation offer some of the most exciting tools for facilitating rapid turnaround time for development and dissemination of products. Simulators— versatile machines that duplicate key characteristics of a situation or system— need to have high quality, accurate, representative information on which to build their databases.

By quantifying tasks in diverse environments (from home to hospice care settings) and mapping these onto the human-robot-environment triad,[12] we will have a better understanding of what it is we need to simulate. Just as in NASA Ames human factors research[14] we are in a dichotomous situation: we must address (1)the problems with today's robotic systems as well as (2)*issues raised by the introduction of new technologies in the future.*[14] Future systems will have novel human interfaces that can only be simulated at the present time. The need to realistically simulate task situations is crucial in order to obtain data that will be useful in designing and programming a robotic aid. A simulation's usefulness depends on the degree to which conditions are accurately modeled. In the future, researchers, designers, developers, and evaluators performing robot task simulation studies will need to better understand the human factors associated with integrated motion so they can apply these principles when designing their systems. Forward thinkers will not simply want to develop robots; they will strive to create beautiful, utilitarian and/or playful systems from the onset. There is no reason why personal or medical care robots have to look or move like a frankensteinian monsters or a gangly arachnoids. Aesthetics can be addressed from the outset and in a novel way. For instance, choreographed motion can be included in the robot task development to produce a graceful, yet utilitarian movement[1].

> *A robotic manipulation aid must be able to interact with its user. It does not operate in the industrial context of repetitious tasks. With an emphasis on flexibility, it must operate in largely unstructured environments requiring effective use of tactile sensation.*[13]

Systems will be designed that will need to fit into differing environments. On the other hand, many of the functions that will need to be performed are similar or identical. For instance, activities of daily living (ADL), such as housekeeping chores, will be needed in all of the health care settings, while specific repetitive tasks such as stacking, sorting and filling meal trays in a hospital or long-term care facility may be unique to that particular category of care environments. While robots may have differing designs, they are likely to employ overlapping technological characteristics. It is important to remember that many of these characteristics are also identical or similar to the needs found in industrial or manufacturing settings as well as space exploration.

The robotic *system* will have to be able to respond to the environment in which it must function. It must also be able to respond to the needs of the end-user. First generation, stationary robotic systems that are finding applications in health and human services have clearly defined niches: cells in which to perform a specific set of tasks. These first systems will primarily be utilized by health care staff, rather than patients. For example, **laboratory robots** (a health and human service applications area) come, at the present time, with their own specifically designed multi-task microcosm (called work stations or work cells) made up of special devices which are developed to assist the robot in its tasks.[10] Because of the scale/size of the robot and its confined position, issues of **safety** can be addressed in a more straightforward manner. As robots become increasingly mobile and able, safety will play an ever-increasing role. Figure 2 displays four *spheres of interaction* in which safety must be considered. **Sensors** that will provide *intelligence* will become critical as the robot moves out of its confined work cell and into the human-environment space. We will either have to utilize environments that are user *friendly* as well as *robot friendly*[10], (robot compatible) or we will find it necessary to adopt or create new environments which are more appropriately designed to accomodate both human and robot limitations. This might be thought of as a **forgiving environment** for both people and robots. Safety features will need to be *built into* the software and hardware configurations from the beginning of the design process.

Safety is and continues to be the number one issue that transcends all other questions. Safety for the human and safety for the robot within their common environment will be of paramount importance in the acceptance and diffusion of robots into health and human service settings. Artificial intelligence applications, especially sensors, will allow us new approaches to addressing this critical challenge. We must apply Asimov's first law of robotics to the design, development, and evaluation of human service robots. This law states that *a robot may not injure a human being, or through inaction, allow a human being to come to harm.*[2]

CONCLUSION

Health and human services robot **applications** provide an excellent bridge between industrial and personal robot manufacturers. From a research and design perspective, institutional health care delivery settings resemble industrial plants in their basic topography. For instance, both settings have long halls and systematic organization of open spaces, workspaces, or aisleways. From the development and dissemination perspective, many of the requirements for advanced technology such as touch sensors or vision are compatible with industrial needs. This overlap in the industrial and personal robot technological requirements may, indeed, reduce the barriers of entry into this market as well as expand the customer base of both industrial and personal robot manufacturers.

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A MICROCOMPUTER-BASED BATTERY MONITOR

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ABSTRACT

A technique has been developed to accurately monitor the capacity of a battery independent of temperature, size, or age of the battery. In the past, these three problems have prevented the development of an accurate and versatile battery monitor. A new technique uses the terminal voltage to measure the percent capacity remaining in a battery, once the battery has remained open circuit for thirty minutes. The Miner's Rule model is used to monitor the capacity removed from a battery during charge or discharge. Using the Miner's Rule model and the open circuit voltage, one can maintain an accurate account of the percent capacity remaining in a battery.

INTRODUCTION

An accurate battery monitor is essential to the electric wheelchair user, just as the fuel gauge is essential to the combustion engine vehicle user. Wheelchair users are frequently discharging wheelchair batteries to dangerously low levels, because they are unaware of the accurate condition of the batteries. This situation shortens the life of the batteries, can damage the cells, and can inhibit the ability of the batteries to be charged to a maximum capacity. These problems are alleviated by the development of the battery monitoring scheme discussed in this paper.

Three popular methods of monitoring the capacity of a battery are[1]:

1.) Measurement of Electrolyte Specific Gravity,
2.) Battery Terminal Voltage,
3.) Battery Modeling.

Each of these techniques have certain restrictions which prevent it from being an accurate and versatile monitoring technique. The monitoring scheme presented has been developed using a combination of the above techniques.

DISCUSSION OF CURRENT TECHNIQUES

The measurement of the electrolyte specific gravity is a direct indicator of the capacity remaining in a battery. A hydrometer is used to measure the specific gravity of the electrolyte. A hydrometer, which is a device which must be manually inserted into a cell of a battery, withdraws a sample of the electrolyte. A special type of float material inside the hydrometer is used to determine the concentration of electrolyte acid to water, thus giving specific gravity. This technique is clearly impractical for the general wheelchair user. Also, some batteries are maintenance free or use a gel type electrolyte which also prevents the use of a hydrometer to determine the capacity of a battery.

Battery terminal voltage also has been shown to be a good indicator of the percent capacity remaining provided the battery remains in an open circuit(no load) state for at least thirty minutes. This restriction makes this technique useless under loading conditions, because of the fact that it does not provide the wheelchair user with an instantaneous indication of the battery capacity.

The technique of battery modeling is usually implemented by means of a microcomputer in which either a look-up table or an equation describing the characteristics of the battery is stored in software. Using the current and discharge time in the model, the microcomputer can perform calculations to determine the capacity remaining in the battery. The problem with this technique is that some models assume that the battery will be under constant current loading conditions. This assumption can cause large errors if there are fluctuating load currents, which is the realistic case for wheelchair applications. Other models try to compensate for fluctuating load currents but do not consider the fact that the characteristics of the battery vary with size and age. Also, after a number of charge/discharge cycles the battery cannot be recharged to one-hundred percent of its initial maximum capacity; thus, causing an error in the model. These errors accumulate with each charge/discharge cycle and will never be corrected.

The technique being developed combines the last two of the above techniques in a way so as to compensate for the restrictions of each. By compensating, an accurate and versatile battery monitor that can adapt to batteries of different size and condition results. Specific gravity measurements were used to verify the technique.

MONITORING TECHNIQUE

In designing the battery monitor, two assumptions were made. The first assumption was that the terminal voltage of a battery is a good indicator of the capacity remaining in the battery, provided the battery has remained in an open circuit state for at least thirty minutes[1]. The second assumption was that this "open circuit voltage" was independent of discharge rate, size, condition, and temperature of the battery.

Various tests have been conducted on two types of deep discharge lead-acid type batteries(Sears 9601, and Donley 22NF) to verify the above assumptions. The first tests were conducted to determine the length of time required for the batteries' terminal voltage to reach an equilibrium state, after discharging the batteries to 90,80,70 percent capacity at a constant current discharge of 5,10,15 amperes. With each ten percent capacity removed, the load current was removed and the open circuit voltage was monitored until an equilibrium state was reached. These tests showed that after thirty minutes, the open circuit voltage was sufficiently constant (see Figure 1). This equilibrium voltage and the length of time to reach it were independent of the size, discharge rate, or condition of the battery tested (see Figure 2). During these tests the batteries electrolyte was maintained at a constant temperature of 85 degrees F. The percent depth of discharge was measured with a hydrometer.

The above tests were also performed with the battery electrolyte temperature stabilized at 40 degrees F. These tests were performed by placing the batteries in ice and leaving them overnight to allow the electrolyte temperature to decrease to 40 degrees F. This electrolyte temperature was maintained throughout the test. Although these tests showed that the specific gravity readings were heavily influenced by temperature, once these readings were compensated for temperature, the open circuit voltage continued to be a good indicator of the capacity remaining (see Figure 3).

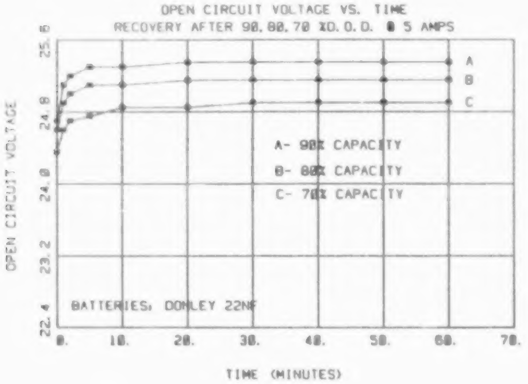

Figure 1.
Recovery Time for Unloaded Voltage

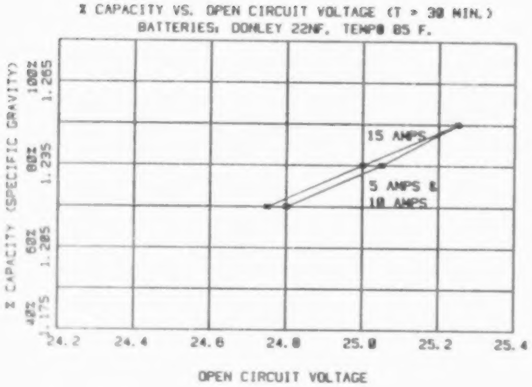

Figure 2.
Dependence of Discharge on O.C. Voltage

The monitoring system will operate in the following manner. At power up, the terminal voltage is read and a corresponding percent capacity is output to a display. The current is then monitored to determine whether or not the batteries are under load. If the batteries are under load, a technique called Miner's Rule is used to monitor the capacity being removed from the battery. The Miner's Rule is an accurate modeling technique for fluctuating load conditions, when given

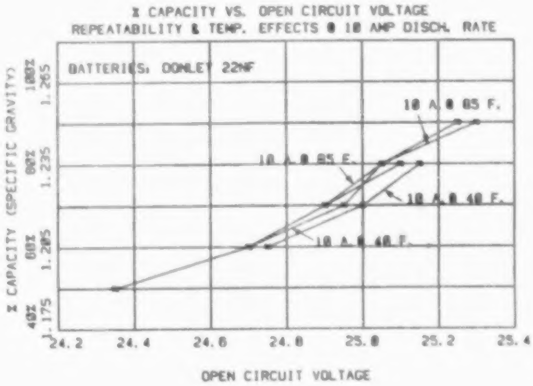

Figure 3.
Dependency of Temperature on O.C. Voltage

an initial capacity and a measurement of the charge or discharge rate[2]. Errors may occur in this modeling technique for several reasons. The first reason is that the same size battery may not be used at all times; thus, causing the initial model to be different than that of the current battery being used. Another reason is that the age of a battery causes differences in its initial characteristics. This is due to the fact that after a battery has been through several charge/discharge cycles, its ability to be recharged to one-hundred percent of its initial maximum capacity decreases. A final reason is that temperature effects can cause changes in the characteristics of a battery. Once the battery has remained unloaded for thirty minutes, the open circuit voltage again will be measured and compared with the calculated value. If any error is present, the model will be corrected so as to maintain an accurate indication of the actual capacity during further use. By using this technique, the monitor will be able to adapt to different size batteries while using nominal initial charastics. This feature allows one to develop an accurate battery monitor that can adapt to different size and condition batteries.

IMPLEMENTATION

A prototype of a battery monitor using the monitoring scheme discussed in this paper has been built around an INTEL 8748 single-chip microcomputer. The terminal voltage is measured directly and the charging or discharging current is measured using a 50 ampere shunt. The shunt and terminal voltages are scaled through an operational amplifier and fed to a 12-bit analog-to-digital converter using an analog multiplexor. The 12-bit converter is implemented using an inexpensive voltage-to-frequency scheme. After appropriate calculations are performed within the microprocomputer, a two-digit LED display presents the percent capacity remaining in the battery. This display is continuously updated as the battery is charged or discharged.

CONCLUSION

Accurately monitoring the capacity of a battery is a difficult problem because of the many different parameters that can affect measurements. The technique presented in this paper is an accurate and inexpensive way of considering these parameters and monitoring the battery capacity accurately. Techniques to optimize the method of adapting the model to the different batteries are presently under development.

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ACKNOWLEDGEMENT

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Alternate Access Methods for Users of Computer-Based Information Systems

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ABSTRACT

Although much rehabilitation information exists in on-line computer-based systems, access to it requires equipment such as a CRT terminal and modem, understanding of how these devices operate, and knowledge of how to communicate requests to the remote computer system. Furthermore, some electronic information systems are inappropriate for visually impaired persons or those with limited physical, financial, or computer abilities. This paper discusses techniques that can overcome these barriers.

NEED

The rehabilitation community with its numerous members, their varied professions and abilities, and geographic dispersion, has many special problems. For example, without continued interaction, organizations and individuals can become isolated from the rest of the community. As a result, rehabilitation research activities and funding are often duplicated, new techniques and devices are not widely disseminated, many experiences of members are not communicated, and potential users of aids may not have a voice in device development.

Computer-based information systems such as NARIC and ABLEDATA have been developed to promote community interaction by addressing users' needs for bibliographic and product-oriented information. Although these systems provide data that are potential sources of solutions, their primary means of access requires personal possession of computer or CRT terminal and modem equipment, knowledge of their use, and expertise with each system's specific command language. The lack of such devices and skills are major barriers preventing the widespread and effective use of these and other on-line information exchange mechanisms.

PROBLEM

Despite the overwhelming need to communicate rehabilitation concerns and solutions, there is currently no universal means to access a forum where such issues can be discussed. Although electronic information systems that remove geographic and schedule barriers to interaction are in operation, they have limitations. Some only allow information retrieval from a large database and are ineffective in the dissemination of users' requests or experienced-based information to other members of the rehabilitation community. Other systems are inaccessible or inapproporiate for those with disabilities. In addition, a high level of proficiency attainable only with considerable training and practice is often required before full advantage can be made of these resources.

SOLUTION

The ideal solution to the information exchange problem is a mechanism that enables rehabilitation community members to make informed decisions based upon an enhanced ability to interact. If a system existed that would allow easy access to its information, the barriers that distance, time, finances, ability, and possession of equipment impose would be reduced, permitting wider participation.

SIGNIFICANCE

Several improvements in the rehabilitation process could result from the employment of new methods which increase user interaction with computer-based information systems. For example, the elimination of the requirement for a modem and CRT terminal would result in a greater subscribership to those on-line systems. Portions of the rehabilitation community that have been excluded from this type of participation due to the burden of acquiring this equipment, the inappropriateness of the display, and/or the effort in learning its operation would then be able to engage in information exchange despite their varying abilities, lack of traditional computer communication equipment, and/or training.

This easier access to computer-based information services could provide a wealth of opportunities to members of the rehabilitation community. Information on devices, job opportunities, travel, and resources could all be more easily obtained if user-appropriate electronic interfaces were available. A system that requires no special equipment for access (other than a telephone), provides interaction between users, is easy to learn and use, and is equally suited for all those interested in rehabilitation issues would provide a significant improvement in information dissemination and informed decision making.

HYPOTHESIS

It is hypothesized that computer-based systems whose information is accessible by a variety of methods can be used by the rehabilitation community to foster increased information exchange and consumer involvement. In particular, telephone accessible on-line computers that employ alternate interfaces would reduce the barriers that presently prevent easy interaction. These new means of retrieving and reporting rehabilitation information could be incorporated into both existing and future system implementations permitting more members of the rehabilitation community to participate in meaningful information exchange.

APPROACH

This project seeks to evaluate alternate methods of access to computer-based information systems. Two classes of interfaces will be addressed, that between the user and the computer system - the user interface and the one between the user and the data - the system interface.

If access to the central information system were modified to accept, in addition to CRT terminal and modem inputs, voice recognition and Touch-Tone inputs and produce synthesized speech or coded Touch-Tone outputs, then these translated user inputs and system outputs would convey the wishes of the user and communicate computer responses in much the same fashion as the customary equipment. Then the possession of, and familiarity with, a standard telephone would be all that would be required. In this manner, those without a CRT terminal and modem could participate in information exchange as well.

One possible access technique translates information destined for hearing-impaired individuals into Touch-Tones. The Echo 2000 [1], Talk-Tone [2], or Teltone T-100 [3] telecommunication devices are portable units designed to decode Touch-Tone signals sent by another user or computer system into displayed letters and numbers. In a similar manner, telephone users could send requests to the remote computer system using the same encoding scheme.

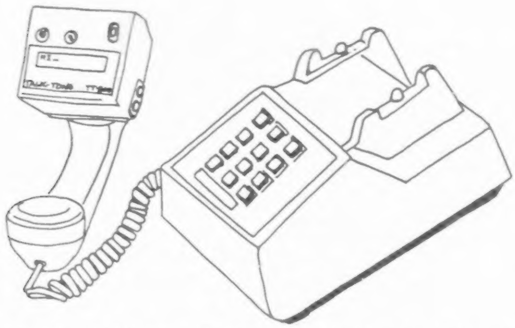

Figure. The Talk-Tone telecommunications device that decodes Touch-Tones and displays alphanumeric information.

The other requirement for successful communication with a remote system is the need to make one's request understandable to that system. Currently each system has its own unique operating language which one must learn. Acquiring this electronic syntax is often difficult because of the multitude of commands and their structure (usually abbreviations), and requires some insight into the workings of the remote system. Since computer technology is responsible for this situation, it might also offer a solution. For example, a software agent can serve as a translator of users' wishes into system commands. This personalized electronic 'tour guide' could assist the user in finding solutions by querying the computer in its own language while responding with a intelligible synthetic voice.

Access schemes drawn from familiar games would also make the retrieval of information easier. For example, the familiar game of Twenty Questions could serve as a model for the interaction between the agent and the user. The agent would 'ask' the user a series of yes/no questions; the user would respond by voice or by pressing the appropriate button on the Touch-Tone telephone. The recognition of the vocal response or decoding of the button-press would give the agent some knowledge of the information the user desires. The agent could then formulate commands in the syntax of the host on-line system to retrieve the requested information. In this example, responses to the twenty questions would select one out of a million items.

Query scenerios based upon television game shows (Let's Make a Deal), computer games (Adventure), and board games (Trivial Pursuit) could be modified for friendly information retrieval purposes. Complex computer command syntax could be replaced by these familiar interactions. In a modified Adventure game, instead of searching for treasure, the user/player would search for rehabilitation information using a mechanism that is both familiar and interesting.

Other interaction possibilities include the manipulation of simulated persons. In this case, the user would be able to select tasks to be performed by these imaginary electronic characters. Observation of the actions of these 'people' as they act out various rehabilitation scenerios could foster information exchange. The user could direct a paraplegic agent/actor to purchase a new wheelchair, participate in therapy, look for housing, or seek a job. This electronic 'soap opera' could provide insight into issues and possible solutions to rehabilitation problems for both individuals with disabilities and health care professionals. A dialog between several simulated characters is also possible given the multi-voice personality of advanced speech synthesizers such as DECtalk. [4]

This project's goal is to design and evaluate alternate methods of access to computer-based information systems. Speech synthesis, Touch-Tone signalling, and speech recognition strategies will be researched as substitutes for the traditional CRT terminal and modem. Several friendly data retrieval schemes will be developed and studied as well. The use of schemes judged to be inexpensive, appropriate, easy to learn, friendly to operate, interesting to use, and efficient in information retrieval could lead to enhanced information exchange that would benefit all those concerned with rehabilitation issues.

CURRENT STATE OF EFFORT

Several user interfaces have been examined. A DECtalk speech synthesizer has been donated and its ability to produce highly intelligible speech has been investigated. Its capability to answer the telephone, speak selected computer text files, and decode user-generated Touch-Tones has been demonstrated with a microcomputer system. Touch-Tone signalling for the benefit of hearing-impaired individuals has also been accomplished. A simple system interface has been implemented with which a user can request and hear abstracts of current projects being undertaken at this Center.

SUMMARY

New and effective ways of exchanging information among members of the rehabilitation community can be identified by evaluating alternate access methods to computer-based information systems. Subsequent implementation of these schemes would improve the quality and quantity of these interactions. A greater subscribership in the information exchange process would result from the elimination of the requirement of a CRT terminal and modem equipment. Portions of the rehabilitation community that have been excluded from participation due to the burden of acquiring this equipment, the inappropriateness of the display, and/or the effort in learning its use would then be able to engage in information exchange despite their varying abilities, lack of traditional computer communication equipment, and/or training. Access to, and use of such an enhanced information system would permit its many users to make educated decisions on issues that affect their work, purchases, prescriptions, and achievement of life's goals.

ACKNOWLEDGEMENTS

Veterans Administration, Jib Ray, Digital Equipment Corporation, Digital Research Incorporated, Palmetto Technologies, Inc., Audiotone, Teltone.

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P1.1

POSTOPERATIVE AND PROSTHETIC MANAGEMENT OF BELOW-KNEE AMPUTEE WITH REMOVABLE RIGID DRESSING AND SCOTCHCAST PREPARATORY PROSTHESIS

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Rehabilitation Institute of Chicago, and
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ABSTRACT

This paper describes two techniques for management of below-knee amputees: 1) The Removable Rigid Dressing for post-operative stump care, and 2) The Scotchcast preparatory prosthesis for early gait training. Our 7 years experience with the Removable Rigid Dressing as a routine procedure and 5 years experience with the Scotchcast prosthesis for nearly 200 patients have shown significant improvement of the total care of below knee amputees.

REMOVABLE RIGID DRESSING

The Removable Rigid Dressing (Figure 1) is a below-knee plaster cast suspended by a stockinette held in place to a supracondylar suspension cuff(6,7).

Making the below-knee plaster cast is slightly different from that of the traditional immediate post-surgical fitting (1) in the following areas: 1). Cotton paddings, 6 layers at the center and tapered to the margins, are used as "spacer" over the bony prominences in order to achieve a controlled pressure relief. 2). The plaster cast is made only to patellar level and wider proximally for easy removal and reapplication of the cast. 3). The cast is held by a combination of suspension stockinette and a Hexcelite supracondylar cuff.

This Rigid Dressing can be applied at the completion of surgery or when the first above-the-knee rigid dressing is removed for wound inspection. When possible, additional tube socks (with elastic band removed) are applied for progressive shrinkage.

Weight bearing exercise starts usually seven to fourteen days after surgery and is determined by the condition of wound healing. The modulation of weight bearing is decided by the stump response.

The Removable Rigid Dressing is worn continuously except for periodic stump observation and hygiene procedures, or when the prosthesis is being worn. The below-knee cast is changed whenever the stump has shrunk to the point at which too many tube socks are being used.

The Removable Rigid Dressing provides immobilization of soft tissue, prevention of trauma, and prevention of edema. Being removable, it allows frequent stump observation without a need for cast-cutting and cast-reapplication (1). More importantly, it permits fast shrinkage by adding tube socks (5,6) and allows observation of the stump response following weight bearing activity. It expedites progressive weight-bearing within the safe tolerance range of the stump. Both undesirable skin breakdown from excessive weight-bearing activity and hesitation in application of early graded weight-bearing stress are minimized.

If there is excessive pressure over the stump, the cast can be softened from outside with a hammer and then pushed from inside for relief. The previous

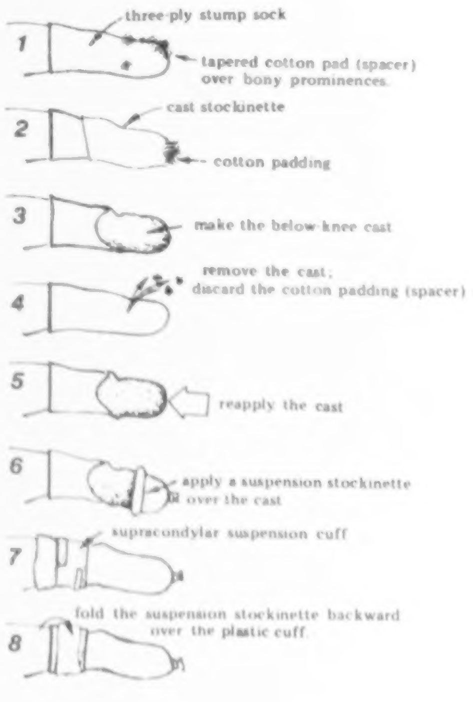
three-ply stump sock
1
tapered cotton pad (spacer) over bony prominences
cast stockinette
2
cotton padding
3
make the below-knee cast
remove the cast; discard the cotton padding (spacer)
4
5
reapply the cast
6
apply a suspension stockinette over the cast
supracondylar suspension cuff
7
fold the suspension stockinette backward over the plastic cuff
8

Figure 1, Making and application of the Removable Rigid Dressing.

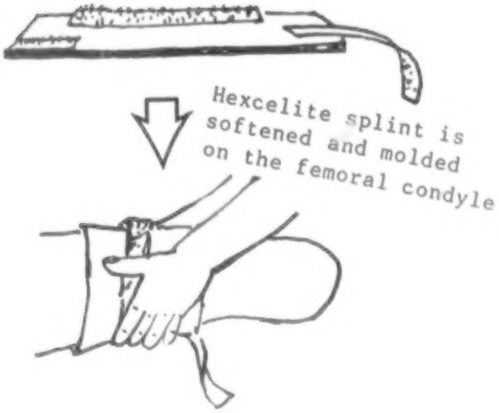
Hexcelite splint is softened and molded on the femoral condyle

Figure 2, Making the Hexcelite suspension cuff.

problems of skin breakdown and distal edema from elastic bandaging have been overcome by this method.

SCOTCHCAST PREPARATORY PROSTHESIS

When the patient can tolerate full weight bearing in the the Removable Rigid Dressing and the stump is no longer edematous, Scotchcast preparatory prosthesis can be made in less than two hours. Making the prosthetic socket is similar to that of Removable Rigid Dressing (6,7) except for the use of Scotchcast casting tape (by 3M Co.).

During casting, moderate pressure is applied to the entire stump and more to the patellar tendon and areas require a relatively higher unit pressure for weight bearing.

After the Scotchcast socket is set, the patient wears tube socks and the Scotchcast socket on the residual limb, and then stands on an adjustable jack to determine the position and comfort fit of the prosthetic socket. Once the most comfortable position of the prosthetic socket is determined, the socket is positioned in the alignment jig, and then joined with the pylon-foot unit using another roll of Scotchcast tape (Figure 5). Finally, attach the supracondylar strap and waist belt for suspension.

The Scotchcast preparatory prosthesis combines the techniques of Lightcast casting (4), the PVC pylon (2) and a new alignment approach (8). The advantages of our approach are light weight, comfortable fitting, rare need for realignment and reduction of fabricating time. This is achieved by (i) direct formation of the socket on the residual limb, (ii) use of wool sock lining as the soft insert, and (iii) precise static alignment of the socket to the prosthetic foot. Since 1979, we have used this system for nearly 200 patients. This preparatory prosthesis has been very practical and effective for early gait training. It has been well received by the patients and the staff.

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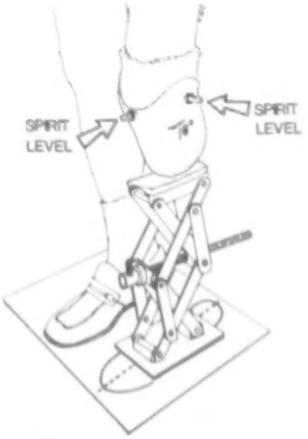

FIGURE 3.
Determine the position (axis and height) of PTB socket by the patient during full weight bearing. Use the anterior and lateral spirit levels to maintain and monitor the mediolateral tilt, flexion of the socket. The height of the socket is monitored by a mark at a specific height, from the ground, 18 inches for a typical male adult.

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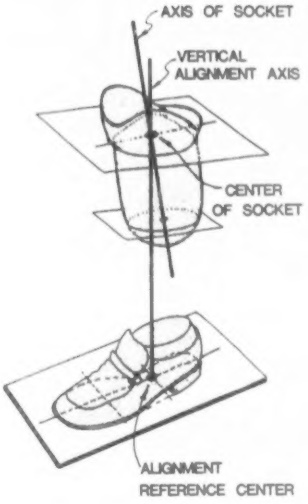
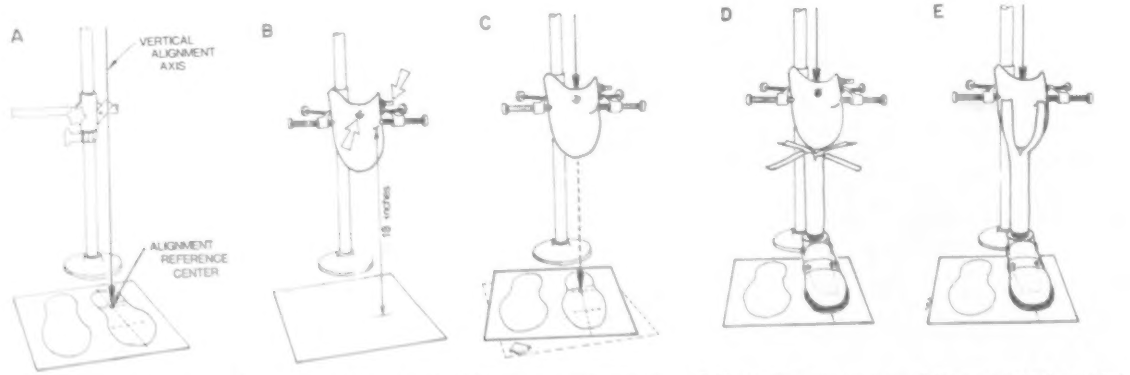


FIGURE 4.
Relationship between the PTB socket and pylon-foot-shoe unit.

Figures 3,4 and 5 are duplicated from Bulletin Of Prosthetics Research,BPR 10-36,Fall 1981.



A. Place the "Alignment Reference Center" of shoe print into the vertical alignment axis (plumb line)

B. Adjust the axis of the socket according to the levels, and the height according to the mark 18 inches from work table top

C. Bring the "center" of PTB socket into alignment axis (pointed by the plumb line) and adjust rotation of prosthetic foot

D. Match pylon-foot-shoe unit to the shoe print on the cardboard

E. Soften and mold the four bars, held by double side mounting tapes, onto the socket

Figure 5, Steps of joining the socket to the pylon-foot-shoe unit.

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EARLY FITTING OF THE AMPUTEE WITH A PLASTIC TEMPORARY ADJUSTABLE BELOW-KNEE PROSTHESIS

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ABSTRACT

Previous experience at the Parkland Hospital and the VA, Dallas has shown that there is a considerable delay in fitting below-knee amputees with a prosthesis, especially if they had their amputations for vascular or diabetic problems. Prior to this study, of the 62 below-knee amputations done at the county hospital, only 21 patients (33%) became prosthetic users. The remainder although walking before their amputation did not regain their ability to walk, mostly due to lack of facilities for early plaster prosthetic application. This study of 42 patients fitted early (2 to 4 weeks) after amputation with a temporary adjustable below-knee prosthesis showed that all regained the ability to walk again, and become definitive prosthetic users. The prosthesis is made from commercially available products and a SACH foot. It is felt to be imminently suitable for institutions where plaster temporary prosthesis are not available.

INTRODUCTION

In a recent study, (2,3) comparing the rehabilitation rate and time of patients in a VA hospital before and after the establishment of an amputee center,using immediate prosthetic fitting, it was shown that all patients who could walk before the occasion of an amputation were effective prosthetic users after the amputation. In contrast, prior to this program only 69% of those who could walk before the surgery became ambulatory with a prosthesis. It was also noted that the average hospitalization for the immediate fit patients was 30.8 days compared to an average of 128 days before the program was started.

The problem of delay in fitting occurs in all hospitals which do not have amputee centers. This is highlighted to a greater degree in public hospitals and VA medical centers where the day to day care is under the direction of residents conforming to traditional speciality organization. As a result the long-term rehabilitation of the patient is frequently forgotten, and attention to prosthetic fitting is not usually considered until wound healing has occurred; by which time the patient has frequently been discharged from the hospital.

One of the major problems has been the provision of a temporary limb within the institutional setting. A recent trend to using temporary below-knee prosthesis made of plaster of Paris has alleviated this problem to some extent. However, since approximately 90% of the amputations in the United States are performed by vascular/general surgeons, who are unfamiliar with the use of plaster of Paris, there is usually a delay in the referral of the amputee for rehabilitation. In practice, the patient is often referred directly by the surgeon to a prosthetist who although competent in the use of plaster and its substitutes, finds it inconvenient to go through the stages of preliminary fitting. He will frequently wait several weeks until adequate wound healing and early stump maturation have occurred. The facilities of the institution must also be considered. The use of plaster temporary prosthesis requires someone with the necessary expertise. Often the institution will discharge the patient with bandages and crutches and will not accept the responsibility of providing a temporary prosthesis. In the medically indigent population no compensation is available from Medicare/Medicaid or community resources for temporary prosthesis.

The purpose of this study was to identify the need for an alternative early fitting prosthesis in settings where appropriate plaster skills are not available. Thus a lightweight, adjustable prosthesis was designed for earlier fitting and ambulation of the below-knee amputee.

BACKGROUND

All of the below-knee amputations performed at Parkland Hospital during the calendar year of 1982 were reviewed. It was found that there were 62 below-knee amputees of which 9 were performed for major trauma and the remainder for vascular disease.

Eight of the nine patients amputated for trauma on the orthopedic service were fitted with plaster temporary prosthesis because of the availability of appropriate skills and personnel on the service. All those fitted ultimately became definitive prosthetic users. However, only 3 of 53 patients amputated on other services were fitted with a temporary prosthesis. Ultimately, 27% of patients (13) on the other surgical services became definitive prosthetic users.

MATERIALS AND METHODS

All consecutive below-knee amputees from orthopedic and general surgery services were evaluated for use of the temporary adjustable below-knee prosthesis. Forty-two patients were fitted early (2 to 4 weeks) after amputation, with the prosthesis. The major criterion for acceptance into the program was adequate wound healing of the stump and recent walking ability prior to the amputation. The cause of amputation in 38 of the patients was for peripheral vascular disease/diabetes, and in the remainder, for trauma. There were 20 male patients of whom 2 were bilateral below-knee amputees and 20 female patients. The age of the patients ranged from 37 years to 72 years with a mean of 53.5 years.

Design and Fabrication

The design parameters of the temporary, adjustable below-knee prosthesis were that it should be adjustable; not made of plaster; and ultimately demonstrate a consistent use of effectiveness for a significant portion of the below-knee amputee population. As simple a design as possible was desirable for reasons of cost effectiveness and low maintenance requirements. This also allowed for continued successful, independent use of the prosthesis by the patient for fitting and training.

In contrast to the traditional plaster temporary prosthesis the adjustable system does not enclose the thigh and allows for a full range of knee motion from the time of fitting. It can also be easily removed for wound inspection and routine hygience care.

The materials used to fabricate the initial prototypes are standard in the prosthetic profession and consists of thermoplastics, urethane foams, and a commercially available pylon system affixed to a SACH foot. The basic concept is one of adjustability to accommodate edema reduction as well as tissue and muscle atrophy via relative positioning of the anterior and posterior sections of the socket. This is accomplished by a series of velcro straps which may be individually adjusted by the patient to accommodate the changing dimensions and contours of the residual limb. The unit is capable of allowing finite alignment changes to enhance the patient's gait and increase energy efficiency. Suspension is provided by a waist belt and billet which attaches to either side of the socket. The patient is fitted with prosthetic socks to act as an interface between the socket and the skin. The prosthesis should be fitted by an appropriately qualified and trained individual.

RESULTS

Of the 42 amputees fitted with the temporary, adjustable system, all became definitive prosthetic users. The average time that the patients used the prosthesis varied from 3 to 9 months. Most of those patients who used the adjustable for a longer period of time had difficulty in obtaining a definitive prosthesis for financial reasons.

In no instances were there any wound breakdowns, or development of localized pressure areas. All patients had received physical therapy prior to and during the use of the prosthesis. In 2 patients who were fitted with the early prototype of the prosthesis, breakage of the junction of the prosthesis to the pylon occurred and they were refitted with a newer model.

DISCUSSION

Under well controlled conditions the majority of below-knee amputees can be expected to be prosthetic users, even though they may be elderly and have dysvascular disease/diabetes. In a consecutive series of dysvascular amputees cared for in an amputee program at Rancho Los Amigos Hospital, 190 below-knee amputees were fitted with temporary plaster devices (6). In this group of patients, 88% were fitted with a prosthesis and used it functionally.

Immediate fitting versus early fitting of the dysvascular amputee probably offers no particular advantage in terms of ultimate prosthetic use, in hospital time, or time to mobilization. In a previous study (5) it has been demonstrated that those with immediate fits as compared to those who only had rigid dressings, spent the same amount of time in hospital. The delay to definitive fitting was the same in both groups. In that same study, those fitted with a plaster pylon achieved a successful definitive prosthetic use at a rate of 74%. Previously the patients at that same institution without early fitting demonstrated a prosthetic fit rate of only 10% (4).

Attempt at creating a nonplaster temporary, adjustable prosthesis have been undertaken in the past. For the above-knee stump, the design constraints are simpler in that the residual limb usually presents no significant bony contours and there is adequate soft tissue cover. Such a prosthesis is available (1) and in a similar study a significantly higher percentage of above-knee amputees became functional users with such an adjustable prosthesis. We are unaware of a comparable series fitted with a lightweight temporary below-knee prosthesis.

The maturation phase of the healing stump has always been a dilemma to the patient, prosthetist and responsible clinician. Contours and dimensions of the residual limb change frequently in the months after surgery. Often the patient to be fitted is debilitated as a result of his disease process and may be underweight at the time of fitting. The situtation may arise where there may be muscle and soft tissue atrophy related to wearing a prosthesis and on the other hand an increase in soft tissue bulk as the patient's general medical condition improves. Thus, fitting a definitive prosthesis in the early months after amputation has the risk of poor fitting due to frequent dimension changes of the residual limb. The below-knee level is an especially difficult problem due to the prominent bony contours that will tolerate poorly the excessive loading which will occur with an ill-fiting socket. In such a situation temporary device provide an opportunity for the patient to become a functional user earlier and to provide an opportunity for earlier tissue maturation.

Thus, we feel that by developing the plastic temporary, adjustable below-knee prosthesis which is reliable and relatively inexpensive it will meet the needs of most below-knee amputees. The patients can be mobilized out of hospital faster, present to the prosthetist sooner with a mature limb and have the opportunity for independent function at a higher percentage than would be expected if no prosthesis were available.

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P1.3

FINGER SWITCH FOR A PORTABLE MICROPROCESSOR SYSTEM TO RESTORE WALKING IN PARAPLEGICS

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ABSTRACT

A ring finger switch was designed and fabricated at the Cleveland Veterans Administration Medical Center for rapid sequential communication between paraplegic subjects and a portable, microprocessor-controlled, 32-channel, electrical stimulator. The ring switch helped to provide a cosmetic and efficient means of controlling the electrical stimulation producing functional walking in these subjects. The ring was fabricated from a thermoplastic material with multilayered, pressure sensitive switches built into the ring and overlayed with a Teflon switch. The devices were hardwired to the microprocessor-based controller with future designs including a radio frequency link.

INTRODUCTION

Ongoing research at the Cleveland Veterans Administration Medical Center has been directed towards achieving functional walking in paralyzed subjects by electrical stimulation (2). The development of a portable, microprocessor-controlled, 32-channel, electrical stimulator (3) allowed programming of several different ambulatory modes such as walking on a flat surface, climbing up stairs and descending stairs. This system required a rapid and relatively simple means of communication between the subject and the microprocessor. The purpose of this communication is to describe the development of a ring finger switch to perform this function. Based on test performance by six male paraplegic subjects, several generations of devices evolved. The design and fabrication of the devices was accomplished through cooperative efforts with the Electronics Design Center, (EDC), at Case Western Reserve University.

BACKGROUND

The paraplegic subjects involved in functional walking, communicate with the portable microprocessor by specific codes. The codes are input by two normally open contact switches which provide essentially two bits of information to the controller. These switches are refered to as the "ready" and "go" switches. A sequence of "ready" and "go" inputs selects an ambulatory mode. Once in the mode for walking on a flat surface, the subject must signal "ready" when prepared and then "go" to begin the stimulation sequence for each step. The subjects had been using a large, hand held, hardwired slide switch to do this switching. As faster gait cycles were achieved, a transition from a choppy gait to one with a smooth shifting between right and left steps required rapid sequential activation of the switches. The difficulty in gripping both a walker and the hand held switch, along with its slow switching cycles, became a limitation to the the subjects and necessitated the design of a finger switch. The aim of this design was to 1) shorten the switching cycle, 2) reduce the effort in controlling the switch, and 3) free the hand for normal function.

MATERIALS AND METHODS

Design Considerations

Location:	A switching device worn on the non-dominant ring finger.
Size/Shape:	A miniaturized device attractive to the subject, resembling a normal ring.
Sensitivity:	A switch sensitivity that ensures reliability during rapid switching yet prevents false switching during accidental contact.
Activation: Mode	Rapid sequential activation of two normally open contact switches. Activation by opposition of the thumb on the lateral aspect of the ring finger.
Microprocessor/switch Interface:	The switches are initially to be hardwired to the microprocessor for evaluation and training. The final device will use a radio frequency link to transmit switching signals to the microprocessor.

Design and Fabrication

The pressure sensitive contact switches designed and fabricated at the EDC consist of successive layers of twenty-mil thick silicone rubber discs, copper conductive tape and a five-mil thick silicone rubber, annular, diaphragm (Fig. I-B). The sensitivity or contact pressure of these switches was a function of the size of the annular diaphragm's inner radius and its thickness (Fig. I-B). Through the subjects' individual evaluations and switching needs, the optimum diapghragm and thus switch sensitivity was chosen for the specific user.

The initial prototype for packaging these switches is illustrated in Fig. I. First, two individual switches were fabricated, each contained within an annular ring approximately, 1/8" in height and 5/16" in diameter. The finger

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ring itself was cut and molded from a commercially available thermoplastic splinting material callled Orthoplast. Under a heat gun, the mold shown in Fig. I was cut and the cavities to hold the switches and lead wires were pressed into the Orthoplast by a separate mold. The piece was then custom formed to the ring finger and was set under cold water. The switches and lead wires were epoxied into the ring and finally the surface of the ring was covered with an elastic or foam adhesive for cosmetic reasons.

After initial subject trials it became clear that Prototype I needed revision. In order to obtain the rapid sequential switching needed for our gait studies, the "ready" and "go" switches must be activated by one sweeping motion of the thumb across the ring finger splint. Prototype II, shown in Fig. II, achieved this ease of switching through a different assembly of the components. The switches were built into a 1/8" acrylic block and overlayed first with a sheet of silicone rubber and then Teflon. The low friction Teflon surface allowed rapid thumb movemnet and the layered switch design allowed replacement of annular diaphragms to obtain optimum switch sensitivity for individual subjects. The acrylic block was again pressed into the orthoplast mold. A drawing of the switch is seen in fig. III. Prototypes I and II were hardwired to the microprocessor-based controller worn on the subject's belt.

RESULTS

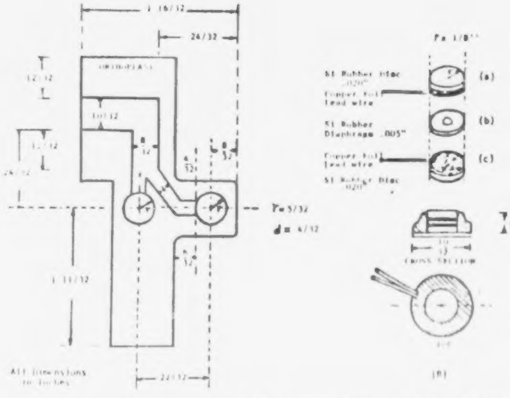
Current subject evaluation includes both positive and negative comments on the design. The subjects feel that the finger ring switch frees their hands for a better grip during walking as well as for normal hand functions while the switch is not in use. Most important, the ease and speed of switching seems to improve their gait speed and control. They also feel that the small size of the switch and the custom fit of the ring is an important cosmetic feature.

Three factors were believed to be responsible for an infrequent yet significant incidence of switch failure during use. First, the switch activation is performed without visual aid. Secondly, the switching surfaces are only 1/4" in diameter. Finally, the high sensitivity built into the switch design was to axial forces only over the surface of the switch. The sensitivity was much less over the rest of the Teflon overlayed switching surface. The requirement of hitting small localized switching surfaces with the sweeping motion of the thumb appeared to be the limiting factor in the switch reliability. Along with better switch reliablity, the subjects expressed a desire for some type of tactile feedback during switch closure. This subject feedback will be included in future designs.

DISCUSSION

Work done in design and modification of these switches has been an iterative process to allow the users to identify and define their needs for communication with the controller. Once a final design is acceptable to the subjects, the ring can be refined by using injection molding processes or having a jeweler adapt a metal ring. The final switch design will also be more quantitatively evaluated for sensitivity, contact bounce, switching life and user reliability.

The problems discussed above concerning the current design suggest a larger, more sensitive, switch surface. Prototype III will investigate various types of membrane switches. The commercial availability and cost will be weighed against the EDC's capability to custom fabricate these switches. The EDC is also working on a radio frequency switch in parallel with this work. Plans are to adapt our micropower,



(A) FIG I PROTOTYPE I SWITCH AND ORTHOPLAST MOLD

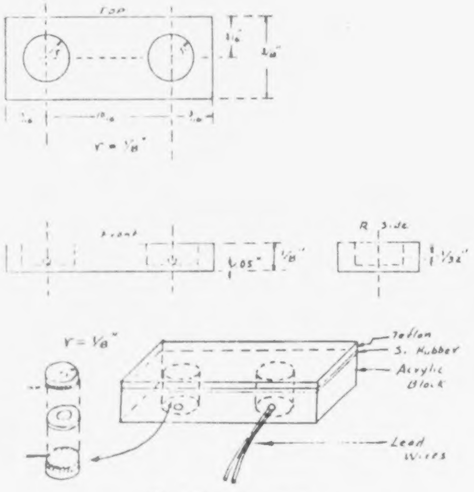


FIG II PROTOTYPE II SWITCH

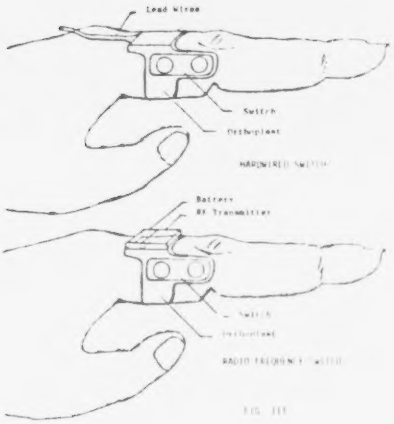


FIG. III

miniature, radio frequency transmitter into the finger switch. M7 K6 modulator/transmitter units, designed and built at the EDC, are being tested for this application (1). The transmitters are constructed in hybrid microcircuit form on an alumina substrate measuring .440 x .485 inches. Figure III suggests the packaging for the transmitter and its battery supply. This switch should totally free the users' hands, leaving him or her with a cosmetic and efficient means of controlling ambulation.

ACKNOWLEDGEMENTS

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A FOOT PRESSURE SENSOR FOR USE IN LOWER EXTREMITY NEUROPROSTHETIC DEVELOPMENT

P1.4

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ABSTRACT

A pressure sensitive device has been developed for use in closed-loop neuroprosthetic systems that accomplish standing and forward motion in paraplegics via electrical stimulation. This sensing device provides pressure distribution information in real time, for use of the controller in the overall system. It consists of four minature pressure transducers embedded within a soft insole, supported by a thin and relatively rigid sole. The performance of this sensor has been evaluated by comparison with measurements obtained using force plates; center-of-pressure measurements have shown excellent correspondence. The performance of this device suggests that foot pressure information sufficiently accurate for control purposes can be obtained using only four transducers for each foot.

INTRODUCTION AND BACKGROUND

A major impediment to the development of practical neuroprosthetic systems using closed-loop-controlled functional electrical stimulation to obtain useful patient locomotion is the unavailability of adequate sensors (1). A closed-loop system for stance has been developed using knee and ankle angle feedback from goniometers (2). For the closed-loop control of gait, however, it appears that real-time measurements of the pressure distribution at the bottom of the foot are needed. The sensor described here was developed to provide this information.

Various methods of measuring the pressure distribution beneath the foot have long been investigated. For this application, a compact sensor is needed to provide feedback to a portable controller box situated at the waist level of the patient. The device is easy to install (an insole that can be worn in regular shoes, each having one multiple connector). The current version of the sensor, as described here was specifically designed for use in the laboratory in the development of closed-loop FES walking systems. For use outside the laboratory we anticipate that,to reduce power consumption, its circuitry will require redesign (through the use of integrated electronics).

MATERIALS AND METHODS

The foot-pressure sensor described here has been obtained by embedding four miniature pressure transducers into the depth of a soft insole supported by a thin and relatively rigid sole. The "model 105" pressure transducer (Precision Measurement Company, Ann Arbor, Michigan) is used. This is a strain-gauge type device where the deflection, due to loading of a diaphragm supported at both extremities, translates into an increase of resistivity of the device above its nominal value. Details of the cicuitry used to achieve good sensitivity and linearity of response with good temperature compensation appear in (3).

An important issue is the mechanical load coupling between the foot and the transducers. The 105 transducer has a thickness of 0.011 in and a diameter of 0.105 in, with one active face which is flat and a dummy surface that is rounded at the edges. It is essential to apply the pressure at the center of the device so as to get maximum diaphragm deflection. The transducer has a hump where the lead wires enter the device (dimensions exaggerated in Fig. 1). If the foot load is applied directly to a supported transducer, the pressure will be concentrated around the hump and, consequently, the sensitive face of the transducer will see less pressure. This effect will be more pronounced if the support surface is very soft, allowing the device to twist and sink into the support material. To reduce this difficulty, the transducer was embedded in an elastomer, with the pressure cell resting on a thin and relatively rigid sole. The transducers were packaged in MDX, an elastomer manufactured by Silastic. Each cell, shown in cross-section in Fig. 1, replaced identically-dimensioned sections of a soft insole (Foam Latex Insoles, #281-6612-155, distr. by F. W. Woolworth Co). Details of this process and the pasting of the hard and soft insoles together (with lead wires inside) are given in (3). A photograph of the sensor is shown in Fig. 2.

The pressure distribution beneath the human foot is well known to be highly nonuniform, both spatially and over time. For example, there are narrow peaks in the pressure function in the heel area at heel-strike, and in the forefoot area at push-off. The nonuniformity of the pressure distribution is due to numerous factors pertaining to the nature of contact between the foot and the shoe, as well as between the shoe and the ground surface.

It is important to realize that for purposes of feedback control the entire pressure distribution is not needed, however, but rather a parameter characterizing it. This is fortunate because the number of transducers requried to sample and reproduce the distribution is unrealistic (economically and with respect to circuit complexity). From a careful consideration of the anatomy of the foot and the control application, the number of transducers needed for each foot can be reduced to four. These are sufficient to obtain good approximation of the center-of-pressure on each foot. Details of the mathematical calculation of the center of pressure from the four transducer measurements are described in (3).

From the skeletal structure of the foot (Fig. 3) one can readily identify bony prominences at the heel and the forefoot, from the medial side view.

These areas are good candidates for transducers since they represent areas of concentration of pressure. Positioning a transducer at the center of the heel is an obvious choice. The location of transducers at the forefoot requires more careful consideration. The metatarsal heads represent the different bony prominences at the forefoot. If a single transducer is placed at the center of the forefoot area, then extreme cases of foot pronation or supination will cause the transducer to lose contact, and therefore introduce a major error in the center of pressure calculation. Consequently a minimum requriement is two transducers placed at the first and fifth metatarsals, respectively. At heel-off, the center of pressure moves to the forefoot area. A fourth transducer is placed in the proximity of the second metatarsal to reduce center-of-pressure estimation error during push-off. A fifth transducer placed under the big toes would further improve resolution, but it was not considered essential.

The insoles are custom-fit for each foot of each patient. The location of the specified bony prominences was determined by careful probing of the foot surface with the fingers (by the electrical engineer assembling the sensor), although other methods such as foot-print stamp pads could be used.

RESULTS

The accuracy of this center-of-pressure sensor was evaluated by comparison with center-of-pressure estimates obtrained using an AMTI six-component Biomechanics force platform (Advanced Mechanical Technology, Inc., Newton, Mass.; Model OR6-5-1.) Analog data was collected from the insole system and the pressure plate through the data acquisition system (DAS) of the MINC lab computer, equipped with a multiplexed A/D converter. Signal processing of this raw data was then carried out on a PDP-11/45 computer, and various graphical displays of the results were obtained using a graphics software package. Details of the software are described in (3).

In Fig. 4 the center-of-pressure estimates obtained from both the insole-sensor and forceplate data are shown for a normal subject during quiet standing. In Figure 5 the center-of pressure location in the x direction is plotted as a function of time for both devices, showing excellent correspondence.

In Figure 6 the center-of-pressure estimates for both devices are shown for a normal subject taking a single step. The agreement is quite good during the middle of the step, with higher errors just after heel-strike and before push-off. The insole detects heel-strike when the force plate does, however.

In Figure 7 the center-of-pressure estimates obtained by both devices are shown for a normal subject during a backward fall. Note that the insole cannot detect the moment the heel leaves the plate.

DISCUSSION

It is important to note that the four transducer arrangement described above has minimum error when all four transducers are touching, as in quiet standing or foot-flat in walking. Maximal errors occur between heel-contact and foot-flat since only the heel transducer is in contact; the center of pressure estimate will thus stay at the transducer location during this time, while the true center is moving forward. During this time interval, however, knowledge of the exact location of the center of pressure is not crucial to controller performance. Note that in Figure 7, although the insole cannot detect the moment that the subject's heel leaves the force plate, the fact that the weight distribution is moving back toward the heel is discernable. We believe that this information should be sufficient to enable a sophisticated controller to detect a potential "heel-off" situation.

ACKNOWLEDGMENTS

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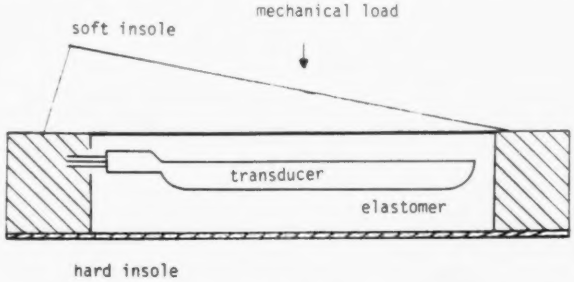


Fig.1 Cross-Section of Pressure Cell

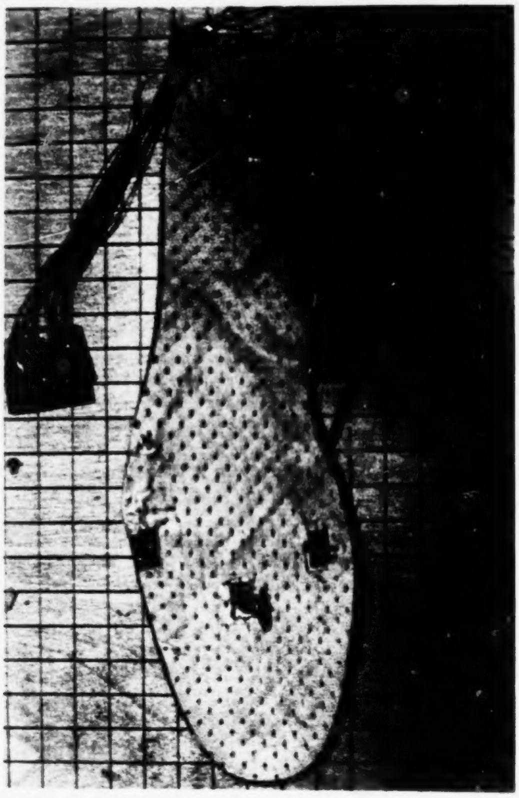
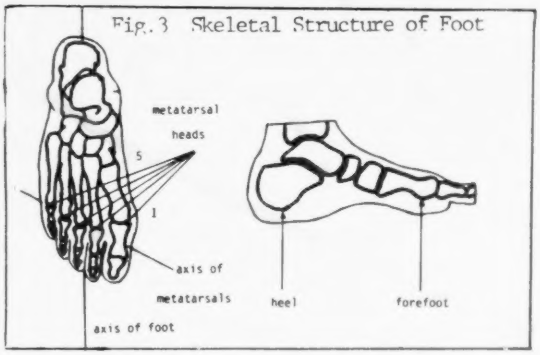


Fig.2 Insole Sensor (top view)

Fig.3 Skeletal Structure of Foot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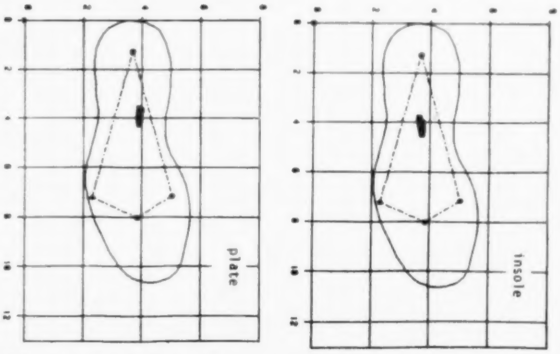
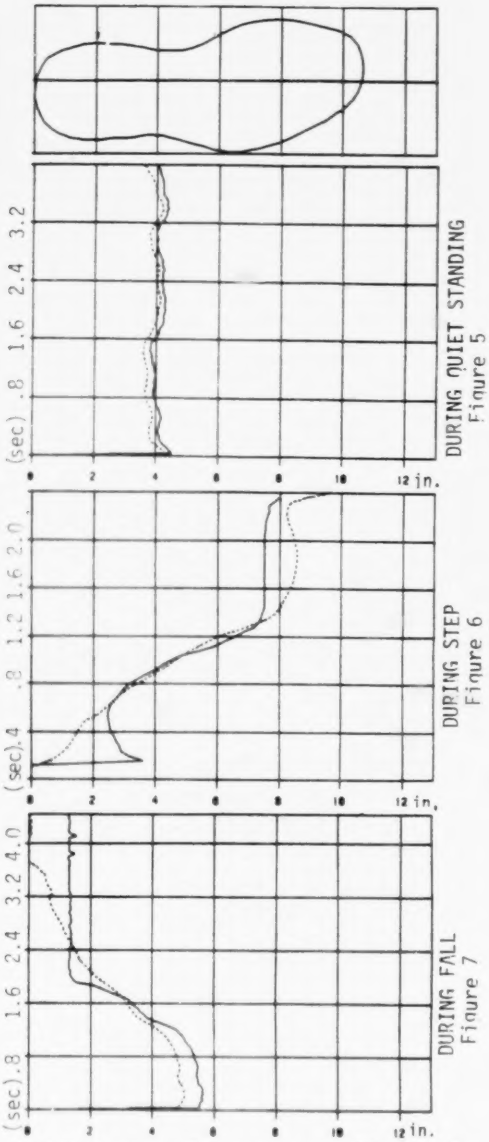


Fig. 4 Center of Pressure Estimates During Quiet Standing from Insole and Forceplate (units inches)



Figs. 5,6,7 Center of Pressure vs. Time during Quiet Standing (5), Step (6) and Fall (7)
Force plate= dashed line
Insole=solid line

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P1.5

MOTONEURON RESPONSE TO INTERMEDIATE AND HIGH FREQUENCY STIMULATION

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ABSTRACT

Studies were conducted on 25 cats to document discharge rates of alpha motoneurons during stimulation at frequencies from 100-4000 pps. In addition, the feasibility of using high frequency pulse trains to block the conduction of action potentials was investigated. Two cuff electrodes were placed around the proximal portion of the left sciatic nerve. Recordings were made from single fibers of the L7 ventral root. When stimulating through the more proximal electrode, discharge rates were equal to or were subharmonics of the stimulation rate up to 1000 pps. Firing often decreased in rate during a two minute run. At 4000 pps, fibers responded briefly at rates of several hundred pps but stopped firing within seconds after stimulus initiation. After cessation of response to the high frequency pulse train, action potentials generated at the more distal electrode did not propagate to the recording electrodes. The "electrical block" so induced could be maintained for up to twenty minutes, and recovery following termination of the pulse train was completed within one second.

INTRODUCTION

Functional electrical stimulation can be used to produce coordinated movements of paralyzed limbs. In so doing, spasticity is often a complicating factor, causing undesired movements that may oppose the induced motion. Available treatments (surgery or chemical injections) are reasonably effective, but they are destructive and eliminate any potential positive use of the spastic muscle. A way of controlling undesired motor activity that is non-destructive and can be initiated and terminated quickly would be extremely useful in many clinical applications.

Studies conducted on the triceps surae of cats by McNeal and Bowman (1) have established that trains of electrical pulses at rates from 100-10,000 pulses per second (pps) caused relaxation of induced contractions. The degree of relaxation was dependent upon pulse amplitude. It was hypothesized that relaxation was due to at least two separate phenomena: 1) rapid depletion of acetylcholine at the neuromuscular junction due to very high discharge rates induced in the motor nerves and 2) a conduction block produced in the nerve at the site of application of the pulse trains. The latter mechanism was reported to only occur at pulse rates in excess of 1000 pps. Similar results have since been reported by Solomonow (2,3).

The purpose of the study reported here was to substantiate or refute the hypothesis presented above and, in the process, document the behavior of single alpha motoneurons in response to intermediate and high-frequency pulse trains.

METHODS

Acute studies were conducted on 25 cats ranging in weight from 3 to 6½ kg. The popliteal space was opened and the common peroneal, tibial and cutaneous branches of the sciatic nerve were cut as far distal as possible and sutured to surrounding fascia to maintain normal tension on the nerve. Cutting these branches eliminated muscular response to electrical stimulation and undesired movement of the preparation. Two electrodes were wrapped around the sciatic nerve; one just distal to the sciatic notch and a second 2 cm proximal to the severed ends. The surface incision was then closed with nylon suture.

Both stimulating and blocking electrodes were identical, consisting of a pair of braided platinum and dacron strands encased in silastic and flattened out at the terminal end to form two bared parallel electrodes 1.5 cm long, 1 mm wide, and 2 mm apart. The electrodes were wrapped loosely around the nerve and sutured closed with the cathode proximal.

An incision was made along the midline of the back, and a laminectomy was performed from L5 to S1. A plexiglass oval ring was sutured inside the outer skin and outside the medial longissimus dorsi muscle. The lumbar cavity was then filled with lightweight mineral oil which had been warmed to body temperature. The dura was cut along the midline and retracted on both sides using small hooked pins which were suspended over the plexiglass ring.

The left root of L7 was carefully isolated and the dorsal and ventral roots separated under oil using a blunt curved dissection tool. The dorsal root was cut and reflected inferiorly. The ventral root was cut at the superior end and placed on a small black plastic tray suspended from the plexiglass ring. The root perineurium was severed using a micro knife and rootlets were carefully dissected using a Zeiss dissection microscope. Individual rootlets were placed on a pair of silver chloridized recording electrodes and tested by stimulating the sciatic nerve at 2 pps and supramaximal intensity. Rootlets were split and retested until it was verified that a single alpha motoneuron was being recorded.

The distal portion of the sciatic nerve was activated with a capacitance-coupled biphasic constant-voltage stimulator operating at 50 pps with a pulse duration of 50 microseconds and an adjustable amplitude. The proximal electrode was connected to a capacitance-coupled biphasic constant-voltage stimulator with an adjustable pulse repetition rate between 1 and 10,000 pps, an adjustable pulse duration between 10 and 1000 microseconds, and an adjustable amplitude.

The recording electrodes were connected to a battery-powered differential amplifier with 100 db common mode rejection, 20 megohms input impedance, 10 kHz frequency response, and a gain of 500. A floating ground reference needle was placed in the longissimus muscle. The nerve recordings and an optically-coupled record of both stimulus pulse trains were fed to a multichannel trigger and a crystal controlled interval counter. Nerve discharge and pulse train intervals were directly fed to a General Automation 1830 data processing system and recorded on digital tape.

RESULTS

Intermediate Pulse Repetition Rates (100-1000 pps)

At a pulse repetition rate of 100 pps, the nerve discharged at the stimulation rate or it did not discharge at all as stimulus amplitude was varied.

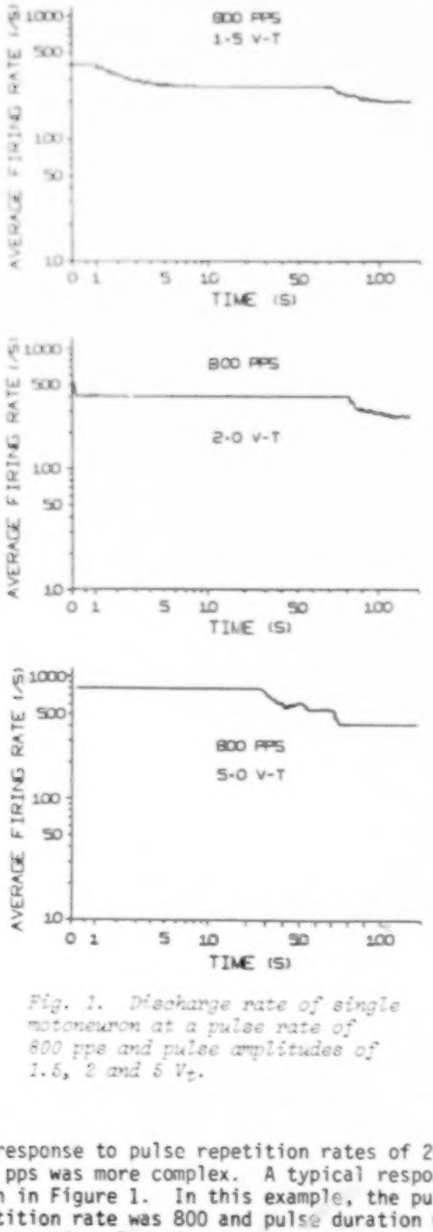

Fig. 1. Discharge rate of single motoneuron at a pulse rate of 800 pps and pulse amplitudes of 1.5, 2 and 5 V_t.

The response to pulse repetition rates of 200 to 1000 pps was more complex. A typical response is shown in Figure 1. In this example, the pulse repetition rate was 800 and pulse duration was 50 microseconds. Records are shown for three stimulus intensities: 1.5, 2 and 5 times voltage threshold (V_t), which is defined as the minimum voltage required to excite the motoneuron at 2 pps. At 1.5 V_t the fiber responded to every second pulse (400/s) for one second, went through a transition period before stabilizing at responding to every third pulse (267/s) from 8 to 70 seconds, and then further slowed its response to every fourth pulse (200/s) by the end of the 2 minute run. As stimulus intensity was increased, the initial discharge rate increased to 800/s, and the transition periods occurred later in time. At 5 V_t a discharge rate of 800/s was maintained for 27 seconds before dropping off to every second pulse.

High Pulse Repetition Rates (2000-10,000 pps)

Responses of a single motoneuron to higher frequency pulse trains are exemplified in Figure 2. In this case, the repetition rate was 4000 pps and pulse duration was again 50 microseconds. When the stimulus was applied at 1.1 V_t, firing began at a peak of 550/s, fell very rapidly (within 0.1 seconds) to 370/s, and continued to fall until firing stopped altogether in just under 10 seconds. Raising the amplitude to 1.4 V_t had the effect of increasing the maximum discharge rate to 620/s and delaying fall-off. This trend continued as amplitude was increased to 1.8 V_t but then reversed as amplitude was increased above 1.8 V_t. At 2.1 V_t the discharge rate fell to 30/s after 100 seconds and at 2.5 V_t response fell to zero in approximately 14 seconds after reaching a maximum discharge rate of only 200/s. Finally at 3 V_t, the nerve stopped firing in approximately 5 seconds after reaching a peak discharge rate of only 180/s.

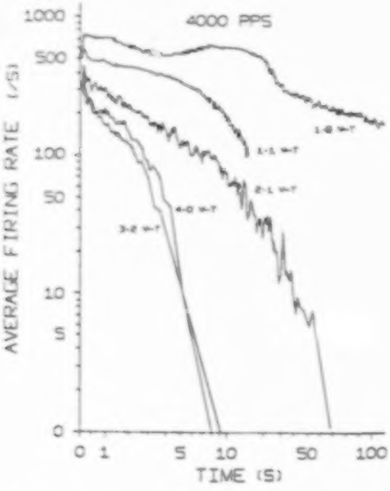

Fig. 2. Discharge rate of single motoneuron at a pulse rate of 4000 pps and various pulse amplitudes.

After demonstrating that an alpha motoneuron stopped responding to pulse trains in excess of 2000 pps, additional trials were conducted to determine whether the motoneuron was able to conduct action potentials initiated at a distal site. In these trials, a supramaximal stimulus at 50 pps was applied at the more distal electrode. The "blocking" signal was applied at the more proximal electrode

5 seconds after initiation of the distal stimulus and was maintained for 90 seconds.

Results for a 4000 pps "blocking" signal are shown in Figure 3. During the first 5 seconds, the discharge rate is 50/s since only the distal stimulus is applied. When a high frequency pulse train at an amplitude of 2 V_t is applied to the proximal electrode (first arrow), the discharge rate jumps to more than 500/s but falls to 50/s within one minute. At 3 V_t, the high rate is maintained throughout most of the time in which the "blocking" signal is on. At 5 V_t, the behavior changes dramatically. There is a very brief increase in the discharge rate, and then the rate drops to 0 about 5 seconds after application of the high-frequency train even though the distal stimulus is maintained. When the "blocking" signal is turned off at 95 seconds, the discharge rate quickly returns to 50/s. (NOTE: Data in Figures 2 and 3 were collected from different motoneurons so a direct comparison cannot be made.) Conduction blocks were maintained in some motoneurons for as long as 20 minutes. In all cases, the discharge rate jumped to 50/s within one second after terminating the "blocking" signal.

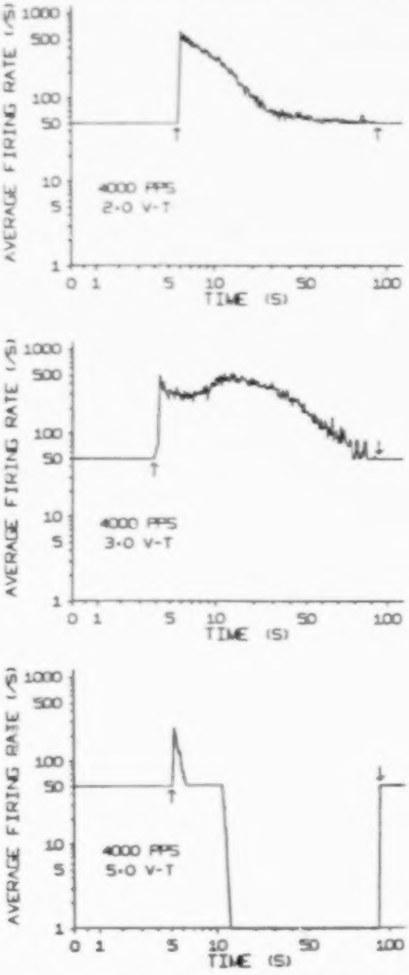

Fig. 3. Discharge rate of single motoneuron with a distal stimulus at 50 pps and a "blocking" signal with a pulse rate of 4000 pps and pulse amplitudes of 2, 3 and 5 V_t.

CONCLUSION

The results presented above support the hypothesis that there were two mechanisms involved in the previous reports on neuromuscular block (1-3). In most cases, high frequency pulse trains cause muscle relaxation by inducing discharge rates in motoneurons far in excess of normal rates. At frequencies greater than 2000 pps, however, a true nerve conduction block can be produced.

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P1.6

ELECTROMYOGRAPHY AND ACCELERATION OF THE TRUNK AS TRIGGER SOURCES FOR IMPLANTABLE GAIT STIMULATION

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ABSTRACT

A microprocessor controlled, implantable gait assist system is being developed at Rancho Los Amigos Rehabilitation Engineering Center. Presently, closed-loop control of stimulation for the various muscle groups of the lower extremities is not attempted. Rather, we are looking for consistent events in the gait cycle which can be detected by implantable transducers and sensors. These events will then be used to trigger stimulation sequences. Traditional external trigger sources, like footswitches, electrogoniometers and handswitches, have been discarded due to their bulk and cumbersomeness. Electromyograms (EMG) and vertical acceleration of the trunk during locomotion have been analyzed. Onset of the EMG of the latissimus dorsi and lateral abdominal muscles occurred consistently just prior to heel-floor contact while a sharp increase in vertical trunk acceleration was detected just after heel contact. Since the EMG can be sensed by tiny wire electrodes and accelerometers are available on a microchip, implantation of these sensors and transducers seems feasible.

INTRODUCTION

In the United States there are 150,000 to 500,000 persons with spinal cord injuries (SCI) and 7,000 to 10,000 new injuries occurring each year. Improved care of acute spinal cord injuries has led to half of these people having incomplete lesions. Ambulation, however, is achieved in only 25% of the total SCI population leaving a large number in need of gait assistance (1). The intent of Functional Electrical Stimulation (FES) gait assist programs at Rancho is to dramatically improve ambulation in SCI persons by using electrical stimulation to actively control lower extremity muscles. All FES gait assist systems require some form of trigger to initiate the stimulating sequence. Currently, most FES gait assist systems are triggered externally by footswitches, electrogoniometers, and/or handswitches. The biggest problem with these external triggering devices is that they are bulky and obtrusive to the user. In addition, equipment abuse leads to frequent malfunction. Using wires to connect these triggers to the stimulator is very unattractive and tends to lessen patient acceptance of the hardware, an often overlooked area of concern. Another problem with using footswitches and electrogoniometers is that they do not give the user adequate voluntary control over the system because they are triggered by the motion of the paralyzed limb. Handswitches, however, do allow the user adequate voluntary control of the system, but having to push a button every step can become very monotonous.

Recently, other triggering methods have been developed that use the motion of the upper extremities and trunk to give the SCI person voluntary control over the system. One such trigger uses strain gauges to detect the forward and backward movement of the shoulders (2). Here again, we have the problem of the trigger device being bulky and obtrusive to the user. Another trigger system uses the above-lesion posture of the SCI person (3). The EMG from the upper trunk is used to map the subjects' posture via a computerized mapping of temporal patterns of EMG. In this system the SCI person must be in the correct posture in order that the desired action be triggered. This system is still being tested, however, the time taken to map the temporal patterns of EMG could limit the walking speed of the subjects.

BACKGROUND

Previous studies of normal gait on vertical acceleration and the EMG of the trunk have revealed a consistent pattern during gait (4,9). From our clinical experience with EMG signals, their most reliable features are the onset and offset times of muscle activity. The amplitude and frequency of EMG signals are unreliable and inconsistent because they vary between each repeated task. Studies of normal gait have shown that the erector spinae muscle group has a very consistent EMG pattern of onset and offset times (4,5). The onset of the erector spinae muscle activity during gait occurred just prior to contralateral heel contact, suggesting the muscles act to resist the passive trunk flexion movement caused during heel contact. In one study of pathological gait, eleven SCI persons showed similar results above the level of lesion (6). This study also showed that the abdominal muscles had a consistent EMG pattern, depending upon posture and level of injury. An advantage of having a trigger just prior to heel contact is that the system has an early warning of this event and therefore time to prepare. Other studies of natural armswing during normal gait have shown that upper extremity muscles also have consistent EMG patterns during gait (7,8). The muscles that showed the most consistent EMG pattern were the anterior, middle and posterior deltoids. Studies of normal gait have also shown that just after heel contact there is a sharp increase in the vertical acceleration (9). Even if the vertical acceleration can be used as a trigger it may not give the SCI subject adequate control over the system. However, one big advantage the vertical acceleration has over footswitches, electrogoniometers and handswitches, is that accelerometers can be manufactured small enough to implant.

Very little work has been done on pathological gait which, obviously, differs greatly from normal gait. One of the primary differences between normal and pathological gait, other than speed, is the use of assistive devices, such as crutches or rollwalkers. SCI persons use these assistive devices for balance and support. The primary muscle (from a kinesiological point of view) that should be active as the crutch or rollwalker is loaded for support, is the latissimus dorsi. Generally the better SCI walkers (walking speeds of 40 meters per minute or greater) use a reciprocating gait and arm swing. Reciprocating gait and arm swing means that the arms and legs move alternately and the arm moves in phase with the contralateral leg. With this type of gait the latissimus dorsi should be active at contralateral heel contact.

Based on findings of the studies discussed above combined with kinesiological considerations we chose to test the following during motion of SCI patients: EMGs from the upper and lower erector spinae, the latissimus dorsi, the lateral abdominals as well as the vertical acceleration of the trunk.

METHODS AND MATERIALS

In order to determine if the trunk EMG and vertical acceleration are reliable triggers, appropriate subjects must be selected. It is the intent of our FES gait assist program to improve the ambulation of SCI walkers, who walk about 20 m/min, to at least 40 m/min. Therefore, by analyzing the EMG patterns and vertical acceleration of SCI persons who walk at least 40 m/min we can determine if, at this speed, these signals provide reliable trigger information.

Eight 50 micron wire electrodes were inserted bilaterally into the latissimus dorsi, the upper and lower erector spinae, and the lateral abdominals. An accelerometer was attached over the lower sacrum. Footswitches and forcecrutches were used to correlate the timing of the muscles in the gait cycle. The subjects, using their own natural style, performed the following tasks:

1) Walking on level ground,
2) Sitting to standing motions,
3) Standing to sitting motions.

All EMG signals were differentially amplified through a Biosentry EMG transmitter and recorded on a Honeywill Visicorder and magnetic tape. The results were digitized and the EMG signals were normalized. The onset and offset times were then tabulated.

RESULTS

Since only a few subjects, at the time of this writing, have been tested, no meaningful statistical data can be presented. However, the results of one representative subject indicate a consistent EMG pattern of some muscles during gait (Figure 1). This subject was an incomplete L2 who had a walking speed of 56.6 m/min and used two forearm crutches with a reciprocating gait and arm swing. The latissimi dorsi showed biphasic activity (two bursts of activity during the gait cycle) at heel contact while the lateral abdominals had consistent activity at ipsilateral heel contact. A sporadic EMG pattern was exhibited by both the upper and lower erector spinae muscles (not shown).

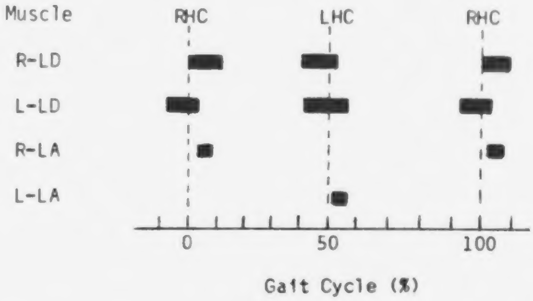

Figure 1. The average duration of electrical activity of trunk muscles during gait in one subject. R-LD, right latissimus dorsi; L-LD, left latissimus dorsi; R-LA, right lateral abdominal; L-LA, left lateral abdominal; RHC, right heel contact; LHC, left heel contact.

During the sitting to standing motion and the standing to sitting motion, the latissimi dorsi showed the most consistent EMG pattern, becoming active almost simultaneously, just before the initiation of the movement.

As expected, the vertical acceleration consistently showed a sharp increase just after heel contact.

DISCUSSION

The initial results indicate that the trunk EMG and vertical acceleration are reliable triggers during gait. The timing of the latissimi dorsi EMGs suggests that the contralateral activity is due to the initial loading of the crutch and the ipsilateral activity is due to the subject pushing of the crutch prior to toe-off to help clear the limb in the swing phase. More subjects will be tested to validate these hypotheses. In further tests some upper extremity muscle will be tested as well, to see if these muscles exhibit useful EMG patterns. In addition to level walking we will investigate the EMG pattern and vertical acceleration during up and down hill gait, and stepping up and down a curb.

The two main advantages to using the trunk and upper extremity EMGs as triggers are: 1) they can be implanted, and 2) they give the SCI person voluntary control over the FES gait assist system. Another possible advantage of using the EMGs as triggers is that the EMG would be adaptive to the subjects' needs. As the subject needs to either walk faster, walk up and down ramps, or step up and down curbs, the EMG pattern would shift, and thereby indicate an appropriate stimulating sequence to the controller. In order to do this, each mode must have a unique trigger sequency associated with it. Certainly, the timing and phase of each trigger will vary between subjects, but using a microprocessor controlled FES gait stimulator will allow one to tailor, by means of software programming, the triggers to that person's gait.

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P1.7

A Closed-Loop Stimulator For Exercizing Paralyzed Muscles

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ABSTRACT

The purpose of this paper is to describe a portable electrical stimulator which can be used to exercise-train muscles of paralyzed limbs by repetitive dynamic contractions. When used in conjunction with a limb position sensor, this stimulator becomes part of a closed-loop, negative feedback system which controls limb position independent of afterload weight. The stimulator output, to surface electrodes, consists of biphasic pulses which are 300 µsec in duration at a frequency of 35 Hz. Output is limited to 150 volts at 150 ma. Circuitry is comprised of three integrated circuits and two transistors, and it can operate on two standard 9-volt batteries. Essentially all of the components are readily available and inexpensive.

INTRODUCTION

Exercising paralyzed muscles of spinal cord injured (SCI) individuals by functional electrical stimulation (FES) has been reported to improve their contraction characteristics (1-7). Although there is no reported return of voluntary function for those whose injuries are complete, there are several suggested benefits which, if substantiated, could justify chronic FES induced exercise. These potential benefits include improved limb integrity, circulation, cardiopulmonary fitness and locomotive capability using FES of the paralyzed leg muscles in conjunction with a leg-propelled vehicle (1,2,3,6,7).

Recently digital computer controlled FES systems which incorporate closed-loop, negative feedback control were shown to offer a high level of paralyzed muscle control (3,6,7). These systems appear to be advantageous over manually controlled stimulators because the level of stimulation is continuously adjusted automatically to permit accomplishment (if it is within the muscle's capability) of the programmed movements, regardless of the afterload force on the muscle. However, disadvantages of such stimulator systems are related to complexity of use and the great expense of the instrumentation. Considering that most exercise training protocols consist of repetitive limb movements through a certain range of motion, it appears that less complex FES systems could be developed to enable performance of these relatively simple exercise tasks.

The purpose of this project was to develop a portable electrical stimulator for exercising the paralyzed limbs of SCI individuals. When used in conjunction with a limb position sensor, this stimulator automatically regulates limb position by negative feedback control. Criteria for the design of this stimulator were that it should: permit smooth, well controlled limb movements; be relatively easy to construct and operate; and be inexpensive.

METHODS

The schematic diagram of the electrical stimulator is illustrated in Fig. 1. Major solid state components are two LM324 quad operational amplifiers (U1, U3), a CMOS 4066 quad bilateral switch (U2), and an IRF333 VMOS output transistor (Q1). The circuit can be powered by two standard 9-volt batteries. Current drain ranges from about 4 ma quiescent to 12 ma at full output.

Ramp Generator for Controlling Limb Position

U1A and U1B comprise a ramp generator for controlling limb position. U1A is a monostable multivibrator which is triggered to the ON state by momentary switch S1, or optionally by an external triggered pulse through an optical isolator (4N33). The duration of the ON state (2-8 sec) is set by R1. The square wave output drives U1B which is an integrator. Output of U1B is a ramp whose rate of rise is adjusted by R2 and rate of fall is adjusted by R3. As shown in Fig. 1, there is a plateau (integrator saturation) between the rising and falling phases. The wave form of this ramp generator corresponds to the intended limb position during a contraction cycle: a rise in limb position; a holding of the limb at a given end-point position; and a return of the limb to the rest position. The end-point positon of the limb is set by R4.

Error Detecting Circuitry

U3A is a differential amplifier which compares the actual limb position with the desired limb position (as indicated by the ramp generator). Potentiometer R6 is the limb position sensor which is placed across the elbow or knee joint. R5 balances U3A for zero output with the limb in the rest position and the ramp generator quiescent. Triggering of the ramp generator provides an error signal from U3A because the desired limb position becomes greater than the actual limb postion. This error signal drives the stimulator for increased output in an attempt to move the limb upward to track the ramp signal. If the actual limb postion exceeds the desired position, U3A output is reversed to decrease stimulator output. Within limits, the greater the tracking error, the greater the signal output (+ or -) from U3A. U3A output is zero with limb tracking of the ramp signal.

Stimulator Drive Circuitry

U3A drives U3B which is the stimulus integrator. U3B serves two purposes: 1- it dampens

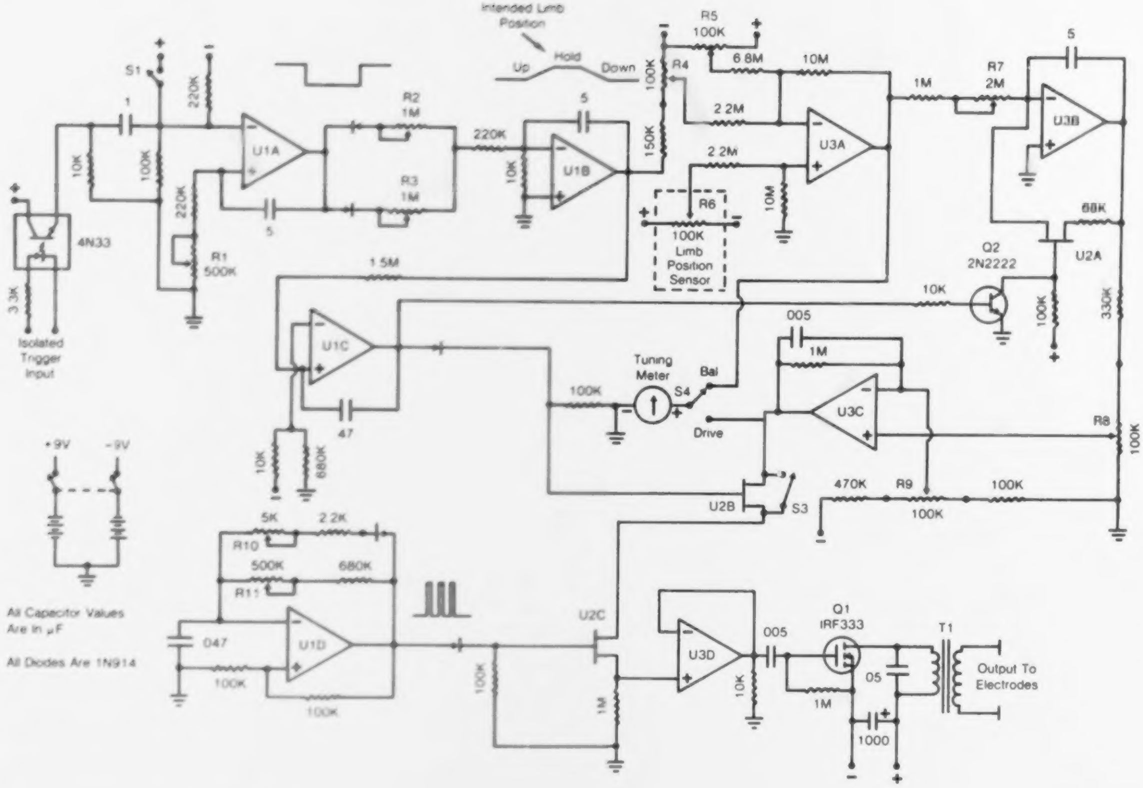
Intended Limb Position
Up
Hold
Down
S1
1
10K
100K
220K
220K
U1A
R2 1M
R3 1M
5
R1 500K
4N33
3.3K
Isolated Trigger Input
220K
10K
U1B
5
R5 100K
R4 100K
150K
6.8M
10M
2.2M
2.2M
10M
U3A
R6 100K Limb Position Sensor
1M
R7 2M
5
U3B
68K
Q2 2N2222
U2A
10K
100K
330K
1.5M
U1C
47
10K
680K
005
1M
Tuning Meter
S4
Bal
Drive
100K
U3C
R8 100K
470K
R9 100K
100K
U2B
S3
+9V
-9V
All Capacitor Values Are in µF
All Diodes Are 1N914
5K
2.2K
R10
500K
680K
R11
047
U1D
100K
100K
U2C
100K
1M
U3D
10K
005
Q1 IRF333
05
1M
1000
T1
Output To Electrodes

Fig. 1. Schematic diagram of the closed-loop electrical stimulator.

the signals from U3A to prevent rapid changes in stimulator output (which could potentially cause injuries); and 2- it keeps the stimulator output at a constant level when the output from U3A is zero (which provides for maintaining appropriate stimulation to hold the limb at the desired end-point position). R7 serves as a stimulator dampening control which is adjusted for smooth movements. U3B drives U3C which is an amplifier with adjustable offset. R8 is used to limit the drive and thus set the maximum output from the stimulator (150 ma across a 1000 ohm resistive load). Offset control R9 is adjusted to provide threshold level of stimulation to the muscle.

Stimulator Output Circuitry

U1D is a clock whose pulses have a width of 300 µsec (set by R10) and frequency of 35 Hz (set by R11). Output pulses drive the control pin of switch U2C to convert the incoming DC stimulator drive voltage from a continuous to a pulsatile signal. U3D serves as a voltage follower to drive the VMOS output transistor (Q1). Output transformer T1 steps the 18 volt from the batteries up to the appropriate output level. It also electrically isolates the stimulator circuitry from the user. Biphasic output pulses are obtained from T1. Surface electrodes (3M Company) are placed over muscle motor points to achieve optimal performance.

Stimulator Control Circuitry

In the resting quiescent state, stimulator output is zero because switch U2A is conducting which short circuits the feedback capacitor of the stimulus integrator (U3B); and switch U2B is not conducting which prevents any stimulus drive voltage from entering switch U2C. With the initiation of a contraction cycle by triggering the ramp generator, comparator U1C switches to the ON state which simultaneously makes switch U2A non-conductive and switch U2B conductive. This provides stimulator drive voltage to U2C, U3D and Q1, which provides a threshold level of stimulator output. Output then increases during the limb rising phase and decreases back to threshold level during the downward phase.

RESULTS AND DISCUSSION

The described stimulator can be manually controlled by depressing S3 and adjusting the output level from threshold to maximal by turning potentiometer R9. It can also be controlled automatically by using a limb position sensor across the joint (knee or elbow) to provide feedback information regarding muscle performance. Fig. 2 illustrates two of these stimulators (one for each limb) in use to provide automatic alternate knee extension exercise. A contraction cycle is typically set for a 70° knee extension over a 3-sec period (concentric contraction), a hold time of 2-sec (isometric contraction), and a return to the rest position over a 3-sec period (eccentric contraction). This exercise device has the limb position sensors (R6) coupled to the shafts of the afterload weight levers.

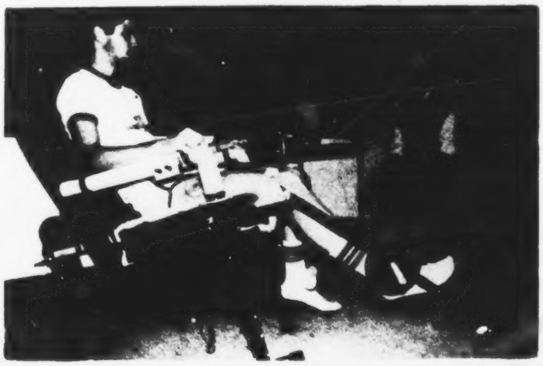

Fig. 2. Closed-loop stimulators in use to provide knee extension exercise. Limb position feedback sensors are coupled to the shaft of the afterload weight levers.

It was found that well controlled smooth contractions could be obtained which are independent of the afterload weight if the muscle is capable of handling the load. With heavier weight and fatigue, the stimulator output automatically increases in an attempt to achieve desired performance. It should be recognized, however, that muscle spasms can interfere with the desired contraction pattern and may impose risk to the user.

Our experience indicates that this described stimulator, which uses inexpensive analog circuitry, provides similar results as does our more complex stimulation system which uses a digital computer for limb position control (3). The analog device is also easier to use since the threshold level, target limb position and hold time at the target angle (as well as other parameters) are set by turning calibrated potentiometers.

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Comparison of Aerobic Metabolism and Cardiopulmonary Responses for Electrically Induced and Voluntary Exercise

P1.8

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ABSTRACT

To compare aerobic metabolism and cardiopulmonary responses for functional electrical stimulation (FES) induced exercise of paralyzed muscle to voluntary (VOL) exercise of non-paralyzed muscle, 4 spinal cord injured (SCI) and 4 able-bodied subjects performed closely matched knee extension tasks. For the SCI subjects, a computer controlled closed-loop electrical stimulator using surface electodes over the quadriceps muscles regulated 4-min bouts of exercise (alternate 7-sec contractions; 2 per min for each leg). Exercise was discontinuous (10-min rest between bouts) and progressive in 5-lb increments (5-20 lb for FES, 5-40 lb for VOL). Mean oxygen uptake and pulmonary ventilation responses were higher for SCI subjects at equal exercise loads. Although resting HR was 11 bpm higher for the SCI subjects, their exercise HR remained below the rest value. In contrast, VOL exercise HR increased above the rest value. These data suggest that FES exercise may be inefficient. This coupled with inappropriate or inadequate cardiovascular responses, could substantially limit the applications of FES exercise.

INTRODUCTION

Research on electrical stimulation of paralyzed muscles is aimed at restoring a variety of functions to spinal cord injured (SCI) individuals (2-9). Major requirements for achieving this aim are that the paralyzed muscle can: be adequately controlled; develop sufficient strength, and be resistant to fatigue. Although much is known about metabolic and cardiopulmonary responses for able-bodied (AB) individuals during voluntary exercise, little data are available concerning these responses for SCI individuals during functional electrical stimulation (FES) exercise.

Two primary reasons that FES exercise of SCI individuals would be expected to elicit different response patterns than for voluntary (VOL) exercise of AB individuals are that sympathetic autonomic nervous system (ANS) pathways are usually interrupted, and that FES exercise is peripherally induced by direct stimulation of the paralyzed muscles. Thus, control of the cardiovascular, pulmonary, thermoregulatory and other systems may be lacking with FES which could reduce exercise capability.

The purpose of this study was to determine aerobic metabolism and cardiopulmonary responses during FES induced exercise of paralyzed muscles and to compare these responses to those for equivalent exercise performed voluntarily by AB individuals.

METHODS

Subjects

Four SCI subjects (2 paraplegics and 2 quadriplegics volunteered for this study. These subjects had recently completed a 12-wk exercise conditioning program which employed FES to strengthen their paralyzed quadriceps muscles (4). Four AB male subjects of a similar mean age volunteered to serve as controls by performing VOL exercise. The protocol and procedures for this study were approved by the Wright State University Institutional Review Board.

Protocol

For the SCI subjects a computer controlled closed-loop electrical stimulator, using surface electodes over motor points of the quadriceps muscle group automatically regulated 4-min bouts of 70° knee extension exercise (alternating 7-sec contractions; 2 per min for each leg) (1,4). The conditioning completed by the SCI subjects employed this exercise rate, and each attained greater than 20 lb load weight capability. AB subjects were paced via tones from the computer and they observed polygraph recordings of their knee extension angles in order to perform the same exercise voluntarily.

Exercise was discontinuous with 10-min rest periods between bouts and progressive with respect to load weight attached to the lower leg (5,10,15,20 lb) for both groups. AB subjects completed additional weights of 25,30,35, and 40 lb.

Physiological Variables

During the 4th min of exercise, steady state oxygen uptake ($\dot{V}O_2$), pulmonary ventilation ($\dot{V}E$) and heart rate (HR) were monitored. $\dot{V}O_2$ (l/min, STPD) and $\dot{V}E$ (l/min, BTPS) were determined by open circuit spirometry. Subjects breathed through a two-way breathing valve. Expired gases were collected during the 4th min of exercise. Aliquots of these gases were analyzed for O_2 and CO_2 concentrations. A cardiotachometer with bipolar EKG chest electrodes allowed continuous monitoring of HR.

RESULTS

All subjects were able to complete the assigned exercise bouts. However, it was noted that the load weight of 20 lb was near the maximal for each of the SCI subjects. In contrast, the AB subjects reported minimal effort to voluntarily complete the tasks even at the 40 lb level.

Figure 1 provides mean $\dot{V}O_2$, $\dot{V}E$ and HR responses, linear regression analysis and correlation coefficients with respect to load weight for both FES and VOL exercise. Although mean resting $\dot{V}O_2$

and $\dot{V}E$ were similar for both groups, HR was 11 beats/min higher for the SCI group. With FES and VOL exercise, $\dot{V}O_2$ and $\dot{V}E$ increased above rest values and tended to be linearly related to the load weight lifted. However, FES responses were higher than VOL responses at given loads. Indeed, $\dot{V}O_2$ and $\dot{V}E$ responses for FES exercise at 20 lb were similar to those for VOL exercise at 35-40 lb. Considering that the SCI group at the 20 lb load was near maximal exercise capability, the FES induced exercise increased $\dot{V}O_2$ and $\dot{V}E$ to only about twice the rest values. Mean HR responses for the SCI subjects at each of the exercise loads were found to be below the rest value; whereas, they were above the rest value for the VOL exercise.

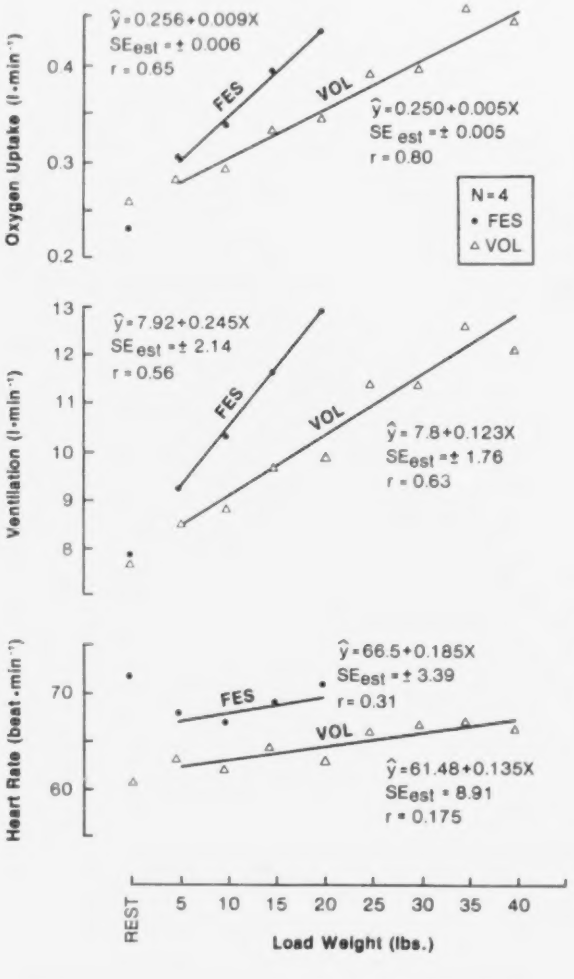

Fig 1. Means, linear regression analysis ± SE_{est} and correlation coefficients for oxygen uptake, pulmonary ventilation and heart rate responses during functional electrical stimulation (FES) and voluntary (VOL) knee extension exercise performed by spinal cord injured and able-bodied individuals, respectively.

DISCUSSION

If FES is to be used as an effective technique to restore function to those paralyzed by SCI and other upper motoneuron dysfunctions, appropriate and adequate organ system (e.g. cardiovascular, pulmonary) adjustments must occur in response to the exercise. With VOL exercise, these responses are normally mediated through the sympathetic division of the ANS. However, FES exercise is peripherally induced which in effect by-passes the ANS. This, coupled with the fact that most spinal cord lesions interrupt sympathetic pathways of the ANS (12) cast doubt as to the ability of SCI individuals to exhibit proper physiological responses to FES exercise (1,2). Such would lead to the early onset of fatigue.

Several studies have shown that FES exercise training is capable of increasing the strength and endurance of paralyzed muscles (3-8). Munsat, et al (10) reported FES induced muscle hypertrophy. Peckham, et al (11) reported FES induced fiber type conversion (fast to slow) and increased levels of oxidative enzymes. These observations are evidence of the ability to restore some of the contraction characteristics to skeletal muscle which had deteriorated because of paralysis. There have also been advancements in controlling paralyzed muscles by computerized closed-looped stimulation systems (4,7,8,9). However, little data have been published concerning the metabolic and cardiopulmonary reponses for FES induced exercise in SCI individuals, and how these responses compare to those normally expected for VOL exercise of AB individuals.

For the present study, the linear increase in $\dot{V}O_2$ with load for FES exercise was expected because aerobic metabolism is locally controlled within skeletal muscle fibers. However, the consistently higher energy expenditure, in comparison to VOL exercise, indicates the inefficiency of this peripherally induced exercise. At approximately the same $\dot{V}O_2$ as the SCI subjects exhibited at 20 lb load, the AB subjects were able to exercise voluntarily at almost twice the load. Several factors may have contributed to this inefficiency for FES exercise. These include: the histochemical characteristics of the paralyzed muscles; inappropriate motor unit recruitment pattern; stimulation of muscle fibers which did not contribute to the performance of the task; and possible initiation of reflex contractions and spasms because of electrical stimulation of sensory afferent fibers (2,10,11).

It was interesting to observe that the $\dot{V}E$ was well regulated (linear) with respects to $\dot{V}O_2$ and load during FES exercise. The exact nature of this control is currently unknown. However, considering the peripheral nature of the FES and the interrupted ANS pathways, it appears to be primarily humoral in nature. The higher $\dot{V}E$ for FES vs VOL exercise suggests greater relative stress for the SCI subjects. It may be that the SCI subjects accumulated a greater concentration of lactic acid in the blood because of more dependence upon anaerobic metabolism.

The absence of elevated HR response during FES exercise is of most concern. This suggests only slight cardiovascular adjustments to this exercise mode. It appears that cardiovascular control is more neurogenic than humoral in nature. Lack of ANS sympathetic outflow would not permit substantial cardioacceleration, and thus decreasing vagal parasympathetic stimulation to the SA node would be the primary means to increase HR. It is feasible that HR lowered below rest value with FES exercise because of the baroreceptor reflex which is initiated by elevated arterial blood pressure. This would result in increased vagal activity to the heart. Inadequate sympathetic outflow may also result in insufficient blood flow to the exercising muscles. This may be due to restricted cardiac output capability and inoperation of the blood shunting mechanism which is necessary to redistribute blood from inactive tissues (e.g. skin, gut) to the active skeletal muscles. Thus, increases in $\dot{V}O_2$ would be primarily accomplished by increases in the arteriovenous O_2 differences rather than elevated tissue blood flow. Poor tissue blood flow will increase dependence upon anaerobic metabolism. The resulting acidosis could rapidly fatigue the exercising muscles. In addition, thermoregulatory responses would most likely be impaired (2,12).

In conclusion, the FES exercise data presented in the study suggest that: maximal capability is severely limited; contractions are inefficient; and that cardiovascular responses may be inadequate or inappropriate. More research is necessary to determine methods of improving FES exercise performance. This would provide more realistic expectations as to capabilities and applications of this rehabilitation technique.

ACKNOWLEDGEMENT

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P2.1

SPECIAL CASES FROM HOSPITAL BASED REHABILITATION ENGINEERING: THREE BED MODIFICATIONS

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INTRODUCTION

The role of a rehabilitation engineer, who is a member of a comprehensive rehabilitation team at an acute care hospital, can vary significantly from case to case. The engineer is often given a referral to evaluate and recommend equipment in the more "standard" areas such as augmentative communication, electric mobility, and seating and postural support. For these, when appropriate, commercially available equipment is usually recommended. Occasionally, unique cases arise where custom hardware is necessary to adequately provide for the patient's needs. Three cases are presented in which Rehabilitation Engineering was consulted for patient bed modifications. After evaluation, investigation of commercial availability, and estimates of cost, it was recommended that a custom adaptation be provided.

CASE I:

SUMMARY

A modification was made to a Roto Rest bed [1] in order to provide alternative positioning of the patient's upper extremities (UE) and prevent further manifestation of UE contractures.

BACKGROUND

Roto Rest beds oscillate along the head to foot axis of the patient. Side bolsters and straps are included to maintain proper positioning of the patient in the bed. Typically these beds are used for acute care of spinal cord injury (SCI) patients for short periods after their injury. Occasionally, a patient must remain on this type of bed for an extended period of time. These beds do not maintain the UE position well, and patients with high quadriplegia and some spasticity who must be on this bed for long periods may develop contractures in their UE.

A patient with C5 quadriplegia, ankylosing spondylitis, some spasticity, and severe UE contractures, was confined to a Roto Rest bed for an extended period of time. His UE contractures included shoulder abduction and internal rotation, and elbow flexion. Proper management of the contractures should include both an aggressive Range of Motion (ROM) program and multiple positioning of the tightened joints. The physical therapist and nursing staff were able to provide ROM therapy two-three times per day; however, this was not adequate to prevent worsening of the contractures. A modification was made to a Roto Rest bed to provide both alternative positioning of the UE and passive ROM of the shoulders.

SYSTEM DESIGN

A sling is suspended from the overhead bar normally used to hold a television. To hold one of the patient's arms in this sling, the elbow is put at 90 degree flexion and the shoulder in horizontal adduction. The extent of shoulder adduction varies with the side to side position of the bed and the height of the arm sling. The opposite arm is positioned along the side of the patient's body. The side bolsters of the Roto Rest bed were modified to accommodate this position. The arms could be positioned alternately.

RESULTS

Because of the pain associated with ROM activities, the patient was only able to wear the arm positioning device for short periods of time (15 minutes to one hour) one or two times per day. Even so, this device greatly enhanced shoulder horizontal adduction and elbow extension, thereby reducing the complications that result from an extended time in this type of bed.

CASE II:

SUMMARY

A device has been added to a standard hospital bed to assist a quadriplegic patient in doing a single independent weight shift while lying in bed.

BACKGROUND

A single 37 year old woman with multiple sclerosis and severe quadriplegia had a history of surgeries and hospitalizations due to the onset of ischemic sores. Because of her medical condition, the patient required a minimum of one weight shift midway through each night's sleep. The patient could not independently shift her position in bed and hired an aid specifically for this task. A system was designed with the goal of permitting independent repositioning.

SYSTEM DESIGN

Pillows are used to prop the patient on her right side. The weight shift is then accomplished by removing these pillows so that the patient rolls from her side to her back. To facilitate rolling, the bed mattress is tilted at approximately a 10 degree angle.

A constant force spring is used to provide the energy necessary for removing the pillows. The spring is held in tension using a ratchet mechanism. This tensing of the spring can be prepared by the patient's night time aid when the patient is put to bed. Midway through the night the

1. Roto Rest Mark 1 Kinetic Treatment Table,
   Manufactured by **Kinetic Concepts**,
   San Antonio, Texas

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patient can then release the ratchet mechanism allowing the pillow to be removed. A force of less than one pound is required to release the ratchet.

RESULTS

Initially, the patient's night time aids had difficulty setting the mechanism properly when they put the patient to bed. However, once this was learned, the system has performed quite reliably.

Since her last hospitalization, several items in the patient's personal care have been altered including: 1) a more rigorous skin inspection, 2) an improved diet, 3) a new mattress, and 4) the weight shifting mechanism described. As a result of these changes, the patient has had no incidence of skin breakdown.

The system has been cost effective in both reducing the amount of personal assistance required and in helping the prevention of the possible onset of pressure sores.

CASE III:

SUMMARY

An apparatus was attached to a bed to help maintain proper leg positioning of a patient who has C5-C6 quadriplegia and severe spasticity.

BACKGROUND

A single 27 year old male suffered from C5-C6 quadriplegia, sensory and motor incomplete, and also had secondary complications of severe spasticity and hyperesthesia throughout the trunk and lower extremities. Spasms occurred regularly throughout the night and caused flexing of the trunk, hips, and knees, and plantar flexion and internal rotation of the feet. Because of the hyperesthesia, it was much too uncomfortable to strap the patient's legs to the bed. Consequently assistance was required regularly throughout the night to straighten the patient's legs and trunk. Rehabilitation Engineering was consulted to provide a system to maintain the legs in the proper position.

SYSTEM DESIGN

The solution consisted of a bungy cord and an AFO worn by the patient. The bungy cord was fastened to the AFO, then strung through pullies at the foot of the bed and finally secured to the bed frame at the head of the bed. This produced an effective spring length of approximately seven feet. The spring force was such that it allowed the lower extremities to flex during a spasm. After the spasm relaxed, the spring force returned the legs to the normal extended position.

RESULTS

The system worked effectively for two to three weeks at which time the patient was discharged to a nursing home. The patient's family set the apparatus up on his bed in the nursing home, and he reportedly used it for a short time. Although the severe spasticity remains, the patient now states he has enough voluntary control and strength in his lower extremities to be able to independently return his legs to extension following a spasm. Therefore, he has discontinued use of the system.

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P2.2

A Home Health Care System for Non-Ambulatory Elderly

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ABSTRACT

How can the functions and services provided by a nursing home or a hospital be brought into the home environment? Our population is getting older and people are living longer. However, the increasing costs of hospital care and nursing home services have made it difficult for elderly persons to afford the medical attention they need. Once admitted to a nursing home, elderly patients often lose the personal contact and attention of their families. They begin to feel lonely and forgotten, and soon lose their incentive to continue therapy and to live. Caring for the elderly at home will not only reduce the high costs of medical care and services, but also will provide them with the family closeness and security that only a "real" home can give.

INTRODUCTION

One of the greatest problems in the care of the elderly in the United States is that this country has not supported and developed home health care services in such a way where the aging individual could comfortably remain at home. What we have forgotten is that Man is often defined by how he lives and where. The home becomes a symbol of security for the old; yet the current trend in the United States is to institutionalize those elderly who have reached the time in their lives where all things become more difficult.

Unfortunately, institutional life tends to encourage unnecessary dependence. A home health care system could provide the impaired with assistance in tasks of daily living, optimal functioning of the individual, and most inportantly, a return to independent living.

BACKGROUND

In the United States there has been a lack of government involvement both in matters of public health and in personal health care. The 1935 Social Security Act, one of the earliest major social welfare programs provided for the development of long-term-care facilities. However, the emphasis in the long-term-care facility is NOT on restoration, rehabilitation, and return to the community; in fact, 80% of the aged who enter these institutions die there rather than in their own homes.

Nursing homes were created to provide for the care of disabled older people, yet they often fail to meet the basic human needs of their patients.

Maslow's hierarchy of needs theorizes that man satisfies his needs in order of priority, and that one must achieve the first level of needs before seeking to satisfy the second level of needs. By comparing the nursing home environment with Maslow's pyramid it is evident that long-term-care facilities are designed and operated to only satisfy the first and second level needs (physiological, security and safety), but are not geared to satisfy the higher needs (social, self-esteem and self-actualization).

There will always be the need for nursing homes. The goal of home health care, however, is to keep the elderly functioning independently in his own home for as long as possible, and to improve the quality of his life.

ANALYSIS

Although not all medical problems can be treated out of the hospital, patients who normally would face long periods of hospitalization such as those with cancer, heart disease, orthopedic problems and stroke, could respond well to home care.

Home care can be beneficial when:
-the home setting can contribute to the patient's recovery;
-the services of a hospital or nursing home are not essential for the patient's needs;
-the patient needs the supportive care of family and friends; and
-rehabilitation is needed for a full recovery.

It is within the personal living space of the home that a home health care system could provide for the following functions:

-sleeping;
-seating;
-work surfaces;
-illumination (both task lighting and ambient room lighting);
-storage and display;
-communication/information/education;
-entertainment/recreation;
-mobility;
-exercise; and
-activities of daily living (ADL).

A home health care system should be designed with the objective of providing an optimum environment for the aged and aging. This environment should promote independence of activity and present challenges to the elderly individual as well as maximize safety and comfort. In designing a home health care system it should be kept in mind that:
-the system must be affordable;
-the system must promote independence on the part of the elderly user;
-the system must have an appropriate level of technology for the user, and
-the system must be sensitive to the needs of the elderly and to changing architecture, product development and economic limitations.

SOLUTION

With the design of the following home health care system, it is possible to create a micro-environment that can support a variety of visual, audio, and tactile stimuli for the enjoyment of one who is bedridden. The end result is a product that can provide the elderly with a safe, efficient system that is compatible within the home environment. Comprised of several components and modules, the system is designed for flexibility in order to provide the user with a wide range of functions and activities.

Fundamental to the system are the following five components:

1. Base
2. Component Support Track
3. Component Support Structure
4. Articulating Bed Surface
5. "Add-On" Components.

The Base is the central component of the system. It houses the hydraulic and electro-mechanical devices necessary to perform the functions of raising, lowering and tilting the system's surface. The base can be lowered to a minimum height of 16" and raised to a maximum height of 27" for nursing purposes. The rubber bellows which surround the lifting mechanism prevent the chances of injury by providing a pinch-free surface.

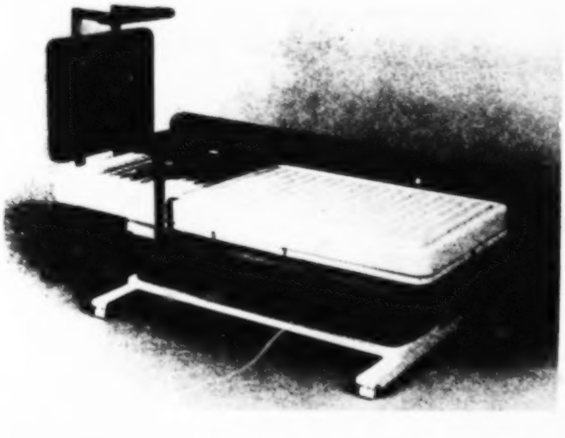
The Component Support Track follows the entire perimeter of the system's base. It is used to make physical and electrical connections between the activity module components and the base. The Component Support Structure consists of a cantilevered bracket which interfaces with the support track. Together, they provide the means by which other system components may be added, depending upon the needs of the particular individual.

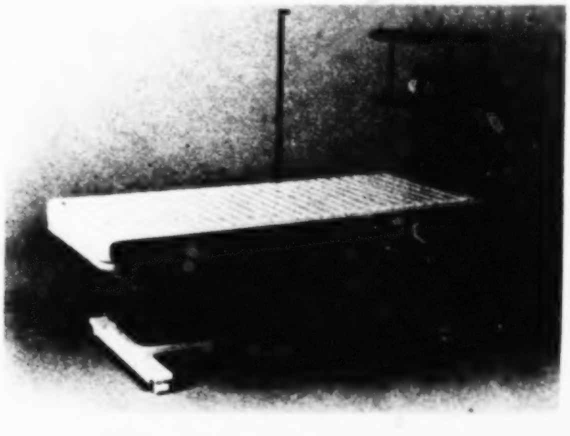
The articulating bed surface provides various positions for sleeping, reading, reclining and elevating. The surface consists of modular gel pads which fit inside a foam outer wall that acts as the perimeter of the surface. The pads remain independent of one another to prevent hammocking and are designed to relieve pressure sores. The pads are impermeable to moisture and can be easily removed for routine cleaning.

The "Add-On" Components consist of the following modules:

The Headboard Module is a padded component that can be adjusted from 24" to 30". The padded surface coupled with the extended height enables an elderly individual to remain seated in an upright position for an extended period of time without creating unnecessary discomfort. This is particularly important for vulnerable areas such as the shoulder blades where the skin is susceptible to breakdown, which ultimately results in pressure sores.

The Worksurface Module is adjustable to various heights and self-locking to various angles. The worksurface can be used for eating, reading, writing crafts and storage; or it can be





flipped-up vertically to expose a lighted mirror which can be used for personal grooming and hygiene. The surface is split into two independent units; both surfaces have a minimum radii to prevent injury during a fall, and a recessed area to contain accidental spills. When the module is not in use, it can be folded down and stored at the side of the bed.

The Information/Entertainment Module consists of a flat-panel screen that is used to communicate audio and visual information to the user. The module also contains speakers and other audio and picture-regulating components. In addition to this, the module is cantilevered and located in a position that provides for optimal unobstructed viewing of audio/visual entertainment and other functions such as in-home banking and shopping.

The Bolster Module provides four significant functions:

1. it provides a firm surface for transferring to or from the bed;
2. it can provide seating for family and friends;
3. it can provide a firm surface for the elderly to sit on while dressing; and
4. when the Bolster Module is not being used for any of the above functions, it can be easily flipped-up and locked in a vertical position to become a protective safety rail.

The Lifting/Exercise Module is interfaced with the headboard component and is designed to perform two important functions. First, as a lifting support bar, the module in its locked position provides an elderly individual with a means of self-help to change positions in bed; to move from the bed to a commode; or to serve as an aid while transferring to and from a wheelchair. Secondly, when the module is released from its locked position, the support bar can be pulled forward and down enabling an individual to perform various exercises. These exercises are designed to strengthen muscles of the upper arm, shoulder and forearm, and to improve flexibility of their respective joints.

The Lighting Module also interfaces with the headboard component to provide both ambient room illumination and task lighting for reading and other work-surface activities. The module is designed to accomodate both standing and wheelchair eye-height levels; and to flatter the elderly individual. Many present lighting conditions create shadows that distort features and over-emphasize health problems.

In certain cases, it may be necessary for the Home Health Care System to adapt to existing medical equipment such as oxygen, kidney dialysis, and I.V. therapies. In the case of I.V. solutions, for example, an I.V. Support Module can be easily interfaced with the Component Support Structure to provide an integrated, adjustable device for administering commercially prepared solutions.

CONCLUSION

One of the most difficult aspects of growing older is the decline in one's ability to function independently and with ease. Therefore, it is important that the environment in which our elderly live provide the necessary means with which this can be accomplished. That environment, however, must not only be functionally fit, it must be emotionally supportive as well.

The Home Health Care System introduced within this paper is just the beginning of what can be done to make the home a place not only where one can be young and grow up, but also a place where one can grow old and still remain.

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A COMMUNICATION DEVICE FOR THE PHYSICALLY DISABLED

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ABSTRACT

A communication device using information theory techniques to minimize user input against speech output is described. Synthesized speech is used as a medium for information output though in practice any form could be used.

INTRODUCTION

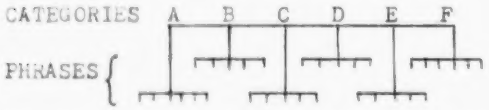
In response to the need of a young cerebral palsied patient with extensive motor dysfunction, a communication device was designed. The need for yet another communication device arose because existing systems were inadequate in one way or another. Keyboard systems are impossible to use for many disabled persons who do not have sufficient motor control, while row-column scanning devices can be slow, cumbersome and expensive. This alternative system approaches a solution to these problems by:

i) using a simple binary code input, compatible with the ability of the patient.
ii) using a coding structure which allows faster access to the most frequently used or most urgently needed phrases.
iii) preserving portability and low cost by using a home computer (Apple IIe) as a work station whereby any number of communication devices may be programmed for the specific and changing needs of an individual under the supervision of a speech therapist.

METHODS

Phrase Menu

The vocabulary of phrases is separated into a number of appropriately selected categories, such as 'Greetings', 'Questions', 'I want...'. One code is used for the selection of the category and a second code for the selection of the phrase.



The use of such a 'two-tier' menu enhances the memorization of access codes. For instance, should the access code for 'Greetings' be 01 the user should quickly become familiar with using this as a prefix to the code for his selected phrase.

In the prototype device, six categories and a total of 50 phrases were used. The limiting factor on the number of phrases is only that of the desired communication rate and the memory recall capacity of the individual (assuming look-up charts are not used after a familiarity with the coding has been acquired).

Coding System Theory

The device takes advantage of a coding structure designed to approach Shannon's Entropy Function (1). Shannon's Entropy Function describes the absolute minimum average number of binary digits required to access information.

In the closed set of data [1, . . . ,n], the entropy function $H(P_1, \ldots, P_n)$ is given by:

$$H(P_1, \ldots, P_n) = K\sum_{i=1}^{n} P_i \log 1/P_i$$

Where:
P_i is the probability of data i occuring.
n is the number of elements in the set (hence $\sum_{i=i}^{n} P_i = 1$).
K and a are arbitrary constants.

When K = 1 and a = 2, $H(P_1, \ldots, P_n)$ is the minimum average number of binary digits needed to access data.

In an effort to approach H, various encoding systems have been devised (2). One of the simplest, and the one used here, is the Shannon-Fano method (3). This method generates variable length codes such that high probability/high priority phrases require short access codes while low probability/low priority phrases require longer access codes. In the prototype device the size of code was limited to a maximum of four bits for either category or phrase selection.

Shannon-Fano encoding. Using this method, each element in a set (categories or phrases) is listed in order of probability of occurrence or pseudo-probability assigned to a phrase of high priority, for instance HELP!).

Example: The set of five phrases is coded as follows.

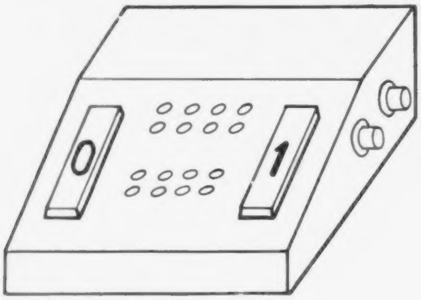
Phrase	P_n		
A_0	1/2	1	1st Division
A_1	1/8	0 1 1	3rd Division
A_2	1/8	0 1 0	2nd Division
A_3	1/8	0 0 1	4th Division
A_4	1/8	0 0 0	

The list is then divided by probability such that the sum of the probabilities above the line and those below are roughly equal. After each division is made, ones are entered above the line and zeros are entered below. The process of dividing is continued until each phrase has a unique code. The high probability phrase A_0 has code 1 while the lower probability phrase A_4 has code 000. The same encoding process is carried out for each set of phrases and for the set of categories.

So that each code is recognized as complete when it has been input, a complex decoding algorithm has been devised using a microprocessor-based system which recognizes each complete code regardless of length.

DEVICE HARDWARE

The prototype device is illustrated below. For evaluation purposes the binary input is made via the mounted keys. In practice switches may be placed anywhere. The display shown is part of a user feedback system described in the next section of this paper.



A Z-80 microprocessor is used to interpret inputs, run the feedback display and output phoneme codes to the SC-01 speech synthesizer.

User Feedback

The patient is provided with feedback by a system of lights. The feedback prompts the user to input the next part of the code and displays the total code input thus far. Two sets of lights are used to achieve this. The upper set provides information concerning category selection and the lower set gives information about the phrase selection.

Category	○ ○ ○ ○	Code input by user
	○ ○ ○ ○	Prompt light to show which part of code is to be input next
Phrase	○ ○ ○ ○	Code input
	○ ○ ○ ○	Prompt

A typical code for a category and phrase might be 01,110. The feedback lights would look like this when 01,11 (all but the last digit) has been input.

Category	○ ● ○ ○
	○ ○ ○ ○
Phrase	● ● ○ ○
	○ ○ ● ○

RESULTS

Dr R Fawcus of the Centre for Clinical Communication Studies, The City University, London tested the device with the patient for whom it was designed with promising results. This has prompted further work on the system which is being undertaken at Tulane University in conjunction with Children's Hospital, New Orleans.

THE FUTURE

A clinic based system is currently being set up, whereby a number of devices may be customized to the particular needs of the patients using the devices. The encoding process, which was done manually for the prototype, is to be automated as is the formation of phoneme strings. This automation will allow rapid reprogramming via an Apple IIe without any need for the therapist to understand the circuitry of the device.

Work on user feedback methods and a more rigorous evaluation will be carried out.

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P2.4

A Proportional Joystick Controller for High Level Quadriplegics

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Abstract

A proportional joystick input control device has been designed to enable a C4, C3 spinal cord injured person (or a person with the equivalent motor function) to control any devices that requires both a proportional input control source and a two switch input source. The total quadriplegic person who has good motor control of the neck and facial musculature and who presently relies on conventional mouthsticks as their controller interface mechanism is now able to have direct input control to environmental control devices, electronic aids, and access to computers utilizing proportional control.

Introduction

The purpose of this project was to develop a means for total quadriparetic individuals to have direct access to computers and other technical aids that require a proportional input control source. The objective was to fit the person with a device that would emulate a conventional proportional joystick controller found as accessories to computer game ports. These also have two momentary switches as additional input signals. Thus, it was important not only to provide a proportional input source but also to provide simultaneous hands free operation of the two single input switches.

There are many recreational software programs and electronics aids which rely on a combination of switch inputs and/or proportional inputs to generate the control command that operates the terminal device. Yet the only controllers that are suitable for quadriplegic use have at most a 5 or 6 microswitch array. (An example would be the TASH ((TM) mini-joystick). This is suitable for operation of a device such as an Atari (TM) game, yet does not provide access to devices that require a proportional input source such as a computer's game port.

Proportional Joystick Prototype I

The original goal for the project was to emulate a conventional proportional joystick in a manner that would allow the quadriplegic to control the device. The design criteria was based on the following:

1. To provide proportional control with activation of two input switches.

2. Cost effectiveness - design the apparatus so the general quadriplegic user could afford to purchase the device.

3. Simplicity in design to facilitate the transfer of technology to others.

The prototype I joystick was based on a modification of a commercially available Taga (TM) joystick which would be used with an Apple II (TM) computer. We chose the Taga joystick because of its availability, low cost, and the ease in which we could modify the proportional controller to suit our needs. We chose the Apple II computer because of its universal acceptance among consumers and the abundance of software available for use.

The two input switch control was designed around a dual pneumatic on/off sensor typically used in switch input technology. We replaced the shaft of the joystick with a hollow brass pipe and fastened a removable plastic dental tube on the end used by the person. The opposite end of the hollow shaft was attached to a dual pneumatic on/off switch assembly. The enclosure to the Taga joystick needed to be enlarged to accommodate the end plastic tubing and pneumatic switch assembly.

The prototype I joystick was fastened to a microphone stand with appropriate adjustable fittings to position the device for the users limited range of motion. The entire assembly was attached to a bedside stand with a removable clamp.

Prototype I - Clinical Usage

Two prototype I joysticks were constructed for use by a C4 level complete spinal cord injured person and a C3 level complete spinal cord injured person. The users tested application programs which required only two switch input, only proportional control, and a combination of both proportional and switch input control. The applications that the prototype I joystick was used for were:

1. Northwestern University keyboard emulator program (1).

2. Software programs using the Adaptive Firmware (TM) as an emulator.

3. Recreational programs requiring proportional control.

4. An input source to a MicroDec (TM for MED product) environmental control unit.

Overall, the prototype I was found to be accepted by the end users and continues to be used on a daily basis. Although the original design criteria was satisfied, there were a new set of design criteria which were developed based upon our observation during clinical usage of the prototype I joystick and from verbal feedback from the two users. The new set of design criteria which would be needed to upgrade the prototype I unit were:

1. The asthetics of the device would have to be enhanced. The prototype I was bulky and non-cosmetic in appearance.

2. The prototype I requires the users to manipulate the joystick through the full mechanical range of the device. The problem with this is that the full mechanical range of the joystick is a function of the length of the shaft (an increase in length requires an increase in the overall amplification of the joystick) and a function of the limited range of motion the individual has. Thus, a variable gain stage would be required to overcome these inherent limitations.

3. The typical self-centering force of commercially available joysticks is often too great for the quadriplegic user. Thus, a softening of the self-centering mechanism would be required.

4. The bulkiness of the prototype I controller impaired the normal field of vision of the user and did not allow for easy access to their mouthsticks.

5. The positioning of the prototype I joystick was awkward due to the many fittings of the apparatus and the change in postural positioning each time the user moved to and from the device.

Proportional Joystick Prototype II.

To overcome these obstacles, a proportional joystick controller based on Hall Effect sensors is being designed. The transducer has the potential to be used in a wide spectrum of applications (figure 1).

The controller consists of four Hall Effect sensors distributed in an orthogonal configuration. The sensors are located within a magnetic field of a magnet mounted on a universal joint. The sensors are differentially connected in pairs. Each pair provides an independent output (x or y) with its own gain control. This characteristic allows the user to have dissimilar ranges of motion in each direction (x or y) and still have a full operating joystick.

Discussion

Since the prototype I controller is based on a modified version of a Taga joystick which has broad market appeal for able bodied user, the cost of modifying the device is moderate. The fabrication procedure can be performed by most rehabilitation engineers. Yet the disadvantages in using the prototype I has led us to consider an alternative means for providing a proportional joystick controller.

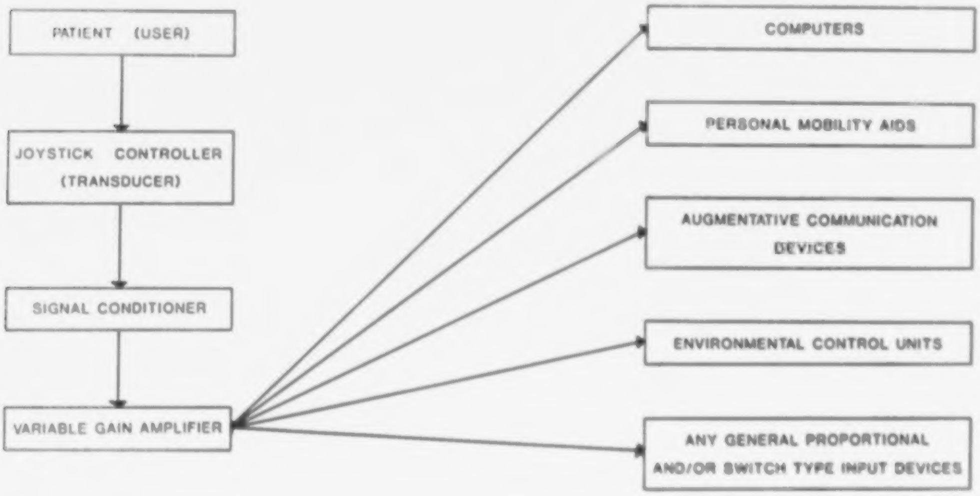
PATIENT (USER)
JOYSTICK CONTROLLER (TRANSDUCER)
SIGNAL CONDITIONER
VARIABLE GAIN AMPLIFIER
COMPUTERS
PERSONAL MOBILITY AIDS
AUGMENTATIVE COMMUNICATION DEVICES
ENVIRONMENTAL CONTROL UNITS
ANY GENERAL PROPORTIONAL AND/OR SWITCH TYPE INPUT DEVICES

Figure 1. Proposed block diagram of the Hall effect joystick interfacing the user to devices.

The Hall effect transducer is presently being refined for clinical testing. One major advantage that the prototype II has over its predecessor is its small size. This allows the device to be placed into a fixture such as a gooseneck (figure 2). The prototype II joystick will enable us to upgrade the performance criteria of the proportional controller unit to fit the needs that were not being met by the prototype I joystick. Another application of the prototype II joystick controller is the potential usefulness that the device may have to researchers in areas of augmentative communication, personal mobility through electric wheelchair control, environmental control, and any areas that could benefit from the power of a miniature proportional controller with a simultaneous two switch input source.

In the future we intend to use the Hall effect transducer in a variety of applications that necessitate a light weight, miniature proportional control source.

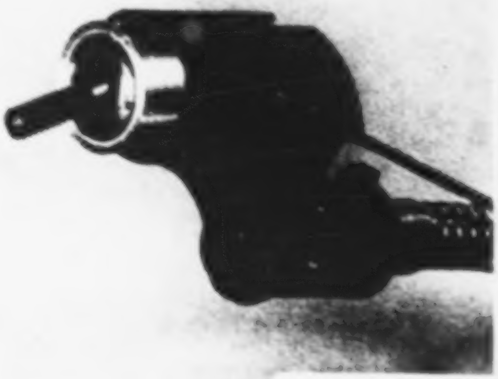

Figure 2. Prototype II joystick in gooseneck fixture. The joystick is based on the Hall effect using a magnetic field and sensors to determine x and y motion.

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Acknowledgements

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P2.5

AVERAGE SELECTION LENGTH AND TIME AS PREDICTORS OF COMMUNICATION RATE

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INTRODUCTION

There have recently been some attempts to devise mathematical expressions which can be used to predict communication rate in text composition and/or speech substitute aids for physically disabled people. These expressions tend to leave out of consideration factors such as user errors, effects of fatigue and learning requirements. As such, they allow the estimation of upper limits on communication rate, and thus have an important role to play in guiding the development of effective systems. In this paper, we present a fundamental (intuitively obvious) equation for communication rate which is general in that it is unaffected by means of indication (scanning, encoding or direct selection), interface medium etc. The relation of this simple formula to expressions presented by other workers, principally Rosen and Goodenough-Trepagnier (1), is described.

The expression involves two quantities only:

L - the average length of a selection in some suitable units (usually characters, but possibly words, etc.)
T - the average time to make a selection.

By a "selection" we mean the elements or units from which the message is composed. Although L and T are interrelated, the relationship is awkward and for the purposes of this paper, we treat them separately. Here, we predict average selection length, L, for a voice-operated text-composition aid which is fully described elsewhere (2). Progress towards prediction of T values for our system, using a set of empirically derived measures, is also presented.

EXPRESSIONS FOR COMMUNICATION RATE

From the above definitions of average selection length and time, communication rate R (under optimal conditions) is clearly given by:

$$R = L / T \qquad (1)$$

Values of both L and T are, of course, required to predict R, and these two variables are interrelated. This is because increasing L generally involves increasing the number of units, N, from which selection is made - which tends to also increase search time, thinking time etc. However, this increased time is strictly due to an increase in N rather than L per se; and it is possible to conceive of circumstances in which N might reduce with greater L (e.g. in going from an inefficient language inventory to a more efficient arrangement.)

In our work, the unit chosen for average length is characters, rather than words, since characters are the more fundamental units. Average word length, in any case, can only be calculated from corpuses which are restricted sub-sets of all possible word occurrences. Thus, R is measured in terms of characters per unit time.

Perhaps the most important existing expression for the prediction of communication rate is that of Rosen and Goodenough-Trepagnier (1). Their equation, justified but not derived, is:

$$\tau = C\ A\ t \qquad (2)$$

where τ = average time per word
C = cost of an inventory (in terms of number of selections required per word)
A = average number of acts per selection
t = average time per act

Apart from using units of words (rather than characters), requiring the introduction of cost, equation (2) is framed in terms of input acts, whereas our equation (1) is based on selections.

We believe equation (1) to be more fundamental than equation (2), since it involves just two very simple terms: and we would argue that the concept of "cost" is not intuitively obvious. In fact, equation (2) can be derived from (1) as we show below.

Average selection time, T, is given by:

$$T = A\ t \qquad (3)$$

Also, cost in equation (2) is related to average selection length in (1) by:

$$C = W / L \qquad (4)$$

where W = average length of a word.

Substituting (3) and (4) into (1) gives:

$$R = W / C\ A\ t$$

Inverting this gives the average time per unit symbol length for a selection:

$$\text{time per unit length} = C\ A\ t / W$$

Thus:

$$\text{time per word} = W\ C\ A\ t / W$$
$$\tau = C\ A\ t$$

Rosen and Goodenough-Trepagnier's expression (2) was originally applied to direct selection and encoding systems only, and these authors believed this to be a limit on the generality of (2). Recently, however, Damper (3) has extended the equation to encompass row-column scanning (when characters are used as selections). Time per word is shown to be :

$$\tau = C\ (A - 1) / r \qquad (5)$$

where A = average number of "acts" per selection. In the scanning case, "acts" includes those occasions where the user does not activate the switch, as the offered symbol is unwanted. Hence, A is the number of binary decisions per symbol - identical to average information in terms of information theory.
r = scanning rate.

Equation (5) was, in fact, derived by extending Rosen and Goodenough-Trepagnier's equation, but equation (1) would have been a better starting point since its generality (i.e. the fact that the expression is independent of the means of indication etc.) is more readily apparent.

AVERAGE SELECTION LENGTH, L

Clearly, then, L is an important characteristic of a language prosthesis. For constant T, the faster of two systems will be that with the higher L value.

L can be predicted with knowledge of language statistics. The average selection length is given by:

$$L = \frac{\sum_{i=1}^{N} n_i l_i}{S} \qquad (6)$$

where l_i = length of the i^{th} element

n_i = number of times i^{th} element was selected

N = size of the inventory

S = total number of "symbols" used, equal to $\sum_{i=1}^{N} n_i$

Average selection length could just as well be defined as:

$$L = \sum_{i=1}^{N} p_i l_i \qquad (7)$$

where P_i = probability of the i^{th} element being chosen, and the identity of (6) and (7) is apparent.

Either equation (6) or (7) could then be used, in conjunction with appropriate statistics, to calculate L. Often, the necessary statistics can be derived from published linguistic data but the derivation is sometimes awkward. To illustrate this, an example estimating average selection length for our voice-input system now follows.

In its original form, the system used direct selection of letters by spoken command to compose text. The inventory was thus 26 letters plus space; and L was equal to 1. In an attempt to increase L, some common words were added to the inventory. The number of words was limited to 10 so as not to compromise recognition accuracy unduly by having too large a recognition vocabulary. The words chosen were those having the highest frequency-length product in the corpus of Carroll et al (4).

This corpus has 5,088,721 words, but the 10 words available for direct selection in our system cover X occurrences. Note that the corpus total does not represent the value S (the total number of symbols used), since apart from the X times whole words were used, the rest of the time was spent selecting single letters. Thus, the number of words left for character-by-character selection is (5,088,721-X). These words contain the following number of characters:

total characters left =

$$(5{,}088{,}721 * 5.74) - \sum_{i=1}^{10} x_i l_i \qquad (8)$$

where 5.74 is the average number of characters per word (including space) (3).

x_i = number of occurrences of i^{th} word

l_i = length of i^{th} word

Thus, the total number of "symbols" (a mixture of whole words and single characters) becomes:

$$S = \text{total characters left} + X \qquad (9)$$

From (8), total characters left equals 25,095,506; and X equals $\sum_{i=1}^{10} x_i$ hence:

S = 26,294,129 symbols

Equation (6) can now be split into:

$$L = \sum_{i=1}^{27} n_i / S + \sum_{j=1}^{10} l_j n_j / S \qquad (10)$$

where the first summation is for letter selection and the second is for whole words. Carroll et al do not give data for n_i and these have to be extrapolated from other tables. The probability of a letter i can be obtained from Baddeley et al (5) and:

$n_i = p_i$ * (total characters left)

Inserting values into (10) yields:

$L = 0.9544 + 0.1584$
$= 1.11$ characters.

Exactly this result is obtained from use of the "mixed mode" equation (i.e. for a mixed inventory of, say, letters and whole words) given in (3). Note, however, that the latter equation is in terms of communication rate rather than average selection length and a slight modification is therefore necessary before it can be used.

To obtain data on communication rate for our voice-recognition system, a relatively informal experiment was conducted, consisting of entering by voice a sample text (from a microcomputer manual) of 89 characters. Communication rate was some 30-40 characters per minute, and the average selection length for the text used was found to be almost exactly 1.1 characters, an increase of 10% over single letter selection. In view of the extremely short text, the extent of the agreement can only be fortuitous. However, the value of the prediction is confirmed.

AVERAGE SELECTION TIME, T

Card et al (6) have presented a model - the so-called keystroke-level model - which predicts the time for a user to complete a routine text-editing task. They decompose the overall task into identifiable unit tasks, and argue that overall time to complete a task having such a sub-structure can be found by summing unit task times for those activities which do not overlap. Users are assumed to be able-bodied and knowledgeable in the task. Essentially the same model is, however, applicable to text composition by physically disabled persons, since selecting a message element involves completion of a number of sub-tasks.

Following Card et al, the time to enter a spoken command is:

$$T_{com} = T_c + T_s + T_{sp} + T_r \qquad (11)$$

where

T_c = cognitive time i.e. the "mental time" needed to think of what is required next.
T_s = search time i.e. the time to search for and locate the required message element.
T_{sp} = speaking time i.e. time to speak a command.
T_r = response time i.e. time required for the recogniser to respond to an input.

In general (although not true for our direct selection system), there will be more than one spoken command required for each selection. For instance, a confirmation input may be required or encoded selection may be used (2). Thus, average selection time can be found by summing average command times, T_{com}.

Examination of the unit tasks in (11) shows that, unlike the average selection length which can be predicted solely from statistics of language usage, average selection time depends on the interface medium and the means of indication. Note that most of the unit tasks are strongly influenced by user abilities with the exception of response time, which depends solely upon the system. (One of the attractions of speech as an interface medium for the physically disabled is that the abilities which assume importance are not motoric.) Unit task times need to be estimated from prior measurements and this causes problems in the case of cognitive time, which is not amenable to direct determination. However, only the non-overlapping component of T_c is required and this can be estimated by subtracting the remaining, easily measurable sub-task times from overall time measured in some representative cases.

Work to allow determination of T for our system is still in progress. In the experiment referred to above, however, average selection time was found to be 1.7 seconds.

A similar performance model has been presented by Gibler and Childress (7). They consider the time to select a language unit in their aid (featuring mechanical switch input and word anticipation) to be the sum of cognitive, visual search, and movement times. The system response time is assumed negligible and cognitive time is ignored due to the difficulty in measuring it. To predict communication rate, these workers combine the selection time measure with a quantity called "text generation efficiency" - the number of inputs required per letter. This is, in essence, merely inversely proportional to our L.

CONCLUSIONS

Equation (1) is a fundamental equation expressing the rate of communication of a text composition aid, made up of two components: average selection length, L, and average selection time, T. The former is a function only of the message elements, and is thus an important design parameter. Using equation (6), L can be estimated from language statistics which can often (but not invariably) be deduced from published data.

Overall rate is also determined by the average selection time, T. T is strongly dependent on the interface design, the user's abilities, and the system response time. Values for these must be measured or deduced before communication rate under optimal conditions can be predicted.

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FOOT SUPPORT WITH 90° KNEE FLEXION ON ADULT WHEELCHAIRS

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ABSTRACT

A wheelchair seating requirement for many developmentally disabled adults is foot support with 90° or more knee flexion. Without custom modification this has not been a commercially available option. Combining a wedge seat with the commercially available adjustable angle footplate has resulted in the desired hip, knee and ankle positioning.

INTRODUCTION

A developmentally disabled adult with abnormal muscle tone presents many concerns relative to wheelchair seating. In working with individuals who exhibit increased extensor tone positioning goals for the lower extremities generally include less than 90° hip flexion; 90° knee flexion or slightly more and slight ankle dorsiflexion. Several developmentally disabled adults with increased extensor tone at Syracuse Developmental Center were poorly positioned in their personal standard wheelchairs with the standard footrests that position the knees in approximately 60° flexion (Fig. 1). The use of these footrests permits the hips to come forward in the wheelchair preventing inhibition of the extensor tone.

MATERIALS AND METHODS

Providing a 4:1 or 5:1 firm wedge seat in the standard wheelchair gave the desired hip flexion. The easiest way to make the modification proved to be using the commercially available adjustable angle footplate (Fig. 2). Use of the camlock footrest bracket was desirable to maintain the "swingaway and detachable" features. By inserting the right adjustable angle footplate/drawbolt adjustment into the left footrest bracket (Fig. 3) and vice versa a level or slightly dorsiflexed footrest can be achieved. If necessary the footplates are trimmed to prevent overlap and holes are drilled for heel loop placement.

RESULTS

This method of modification has been successful for 5 developmentally disabled adults who have increased extensor tone or knee flexion contractures and lower leg lengths 14 inches or less. They are using a 4:1 or 5:1 wedge seat in their standard wheelchairs with the modified footrest (Fig. 4).

DISCUSSION

The Adaptive Equipment Shop at Syracuse Develop-

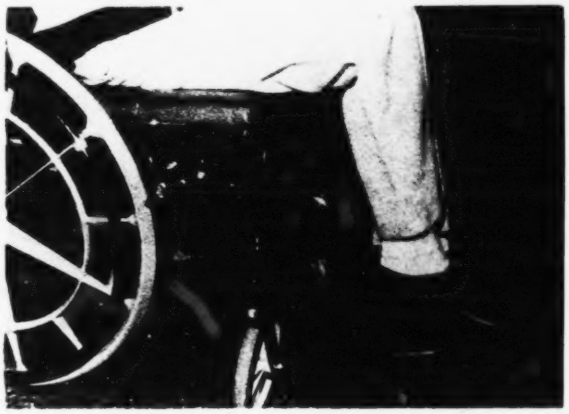

Fig. 1. Position of hips and knees with standard footrest.

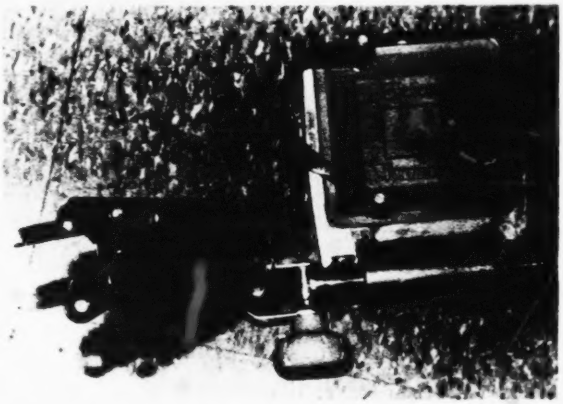

Fig. 2. Adjustable angle footplate.

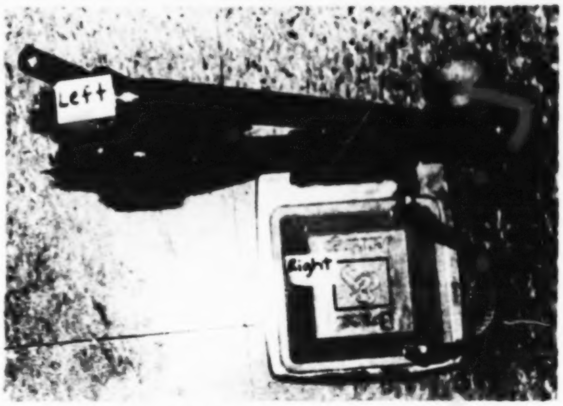

Fig. 3. Right adjustable angle footplate into Left footrest bracket.

Fig. 4. Wedge seat combined with modified footrest.

mental Center has successfully provided optimum lower extremity positioning in a standard wheelchair for several adults without expensive factory custom modifications. As a result of these experiences and the limitations posed by the 8 inch caster wheel our modifications now include changing to a 5 inch caster wheel with adapter. This enables us to accommodate lower leg lengths greater than 14 inches.

ACKNOWLEDGMENTS

I express my appreciation to the administrative staff of Syracuse Developmental Center for establishing and supporting the Adaptive Equipment Shop and to Mr. William Dick, the Adaptive Equipment Specialist, for his outstanding skill in fabricating our designs. Thanks also to Mr. Dave Blatchley for the photography and Ms. Ellen Gooley for her patience in typing.

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RIGID PELVIC RESTRAINT

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ABSTRACT

A rigid pelvic band has proved effective in maintaining pelvic positioning for people with severe athetoid cerebral palsy. The Rigid Pelvic Restraint (RPR) when used in conjunction with a firm intimately contoured seating system, markedly improves sitting position, stability and comfort. This device is now being developed for clinical use for the University of Tennessee Rehabilitation Engineering Program's Service Program (UTREP-SP).

INTRODUCTION - PROBLEM STATEMENT

Maintaining a proper and firmly controlled pelvic position is very important in properly seating a person with severe athetoid cerebral palsy (C.P.). Without such control these people are unable to maintain a comfortable and functional seated posture. Their often wildly fluctuating movement patterns, from flexion to total extension, gradually cause their pelvis to rotate posteriorly and slide out from under even the tightest lap belt. The lap belt, no longer snugly under the Anterior Superior Illiac Spines (ASIS) now constricts the soft tissues of the abdomen. Because the resulting sacral sitting posture reduces apparent shoulder and head height, the back component of their seating system and headrest no longer fit well. At least two of the institutional clients seen by the UTREP-SP have slid down far enough in their seating systems to be choked by their chest belts. Lower extremity position is also poorly controlled. Typically sacral sitters' distal thighs are poorly supported and the seat appears to be many inches too short for them (Figure 1). The unsupported weight of the thighs acts to leverage their pelvis further forward under the lap belt. The resulting semi-reclined posture is a precarious one which triggers an increase of athetoid movement. So seated these people have little chance to develop and use the volitional control they do have. They are more likely to spend a good deal of time in bed and thus be excluded from participation in peer group activities. This is especially discouraging for the athetoid C.P. population who seem to be as a group, more intelligent and motivated than the other severely involved C.P.'s the UTREP-SP staff have worked with.

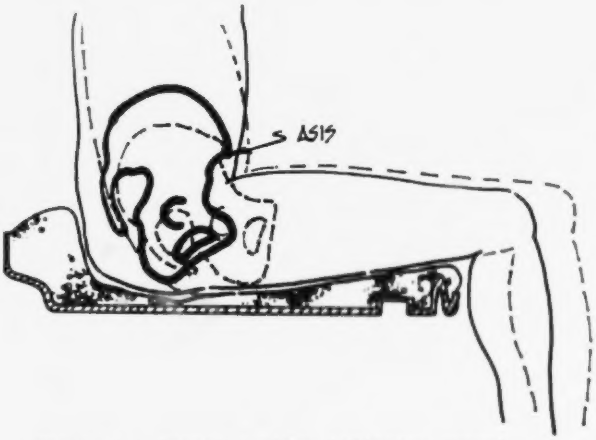

Figure 1 - Dotted outline indicates sacral sitting posture.

CONCEPT DEVELOPMENT

The idea for a RPR has its foundation in proven seating design principles. For nine years the UTREP-SP staff have found that the positive control provided by a firm seating surface is essential for overall postural stability. Standard wheelchair sling seats have been discarded in favor of more rigid structures (1). I propose that the same principles which apply to the seating control elements below the pelvis, the seating surface, also apply to the restraining devices above the pelvis, the lap restraint. If a fabric sling seat poorly controls pelvic stability, a fabric lap belt is also unstable. If a firm seating surface promotes pelvic stability from below so to must a rigid restraint from above.

I made for myself an RPR using polyethelene foam and 1/4" ABS plastic. This device proved much better than a tight lap belt in preventing side-to-side pelvic tilt. It also seemed more comfortable. The restraint forces were better distributed than with a lap belt.

CLINICAL TRIALS

Working in conjunction with the UTREP-SP therapists, I completed the following six RPR fittings.

Andy B.

Andy is a gaingly 18 year old C.P. athetoid living at a local developmental center. He is bright and communicates by pointing to a communication board atop his wheelchair lap tray. He cannot propel his manual wheelchair and is totally dependent for activities of daily living. Since 1977 the service staff had had only limited success in properly seating Andy. His constant movement and particularly strong extensor thrust rendered three successive seating systems only marginally effective. Andy's caregivers reported an incident in which he slid down in his chair nearly choking on the chest belt. After this his lap belt was very tightly fastened. Although this prevented further such incidents the belt began to cause skin breakdown on his pelvis in the areas around his ASIS. This problem confined Andy to bed preventing him from attending classes. In 1982 Andy's seating system was fit with an RPR similar to the one I had made for myself. His seating consisted of a modular plastic back and a firm foam and plywood seat mounted in a wheelchair. The back was perpendicular to the ground plane; the seat was wedged up at a 30 degree angle.

Fabrication of the RPR consisted of molding a 2" wide band of 1/4" thick ABS plastic to roughly fit Andy's pelvis. Contoured pieces of 1/2" polyethelene foam (Evazote) were added to provide a form-fitting resilient surface. Bilateral auto belt buckles and tabs were used for secure attachment to the seating component plywood substructure.

An immediate advantage noted by Andy's therapist was the fact that it was impossible to latch the device unless he was placed fully back into the seating system. In the past his caregivers had carelessly fastened the "D" ring velcro lap belt when he was in an extended posture. Surprisingly the RPR held Andy firmly without causing skin breakdown or discomfort.

In December 1984 Andy received a new Bead System, a custom contoured seating system and a new modular RPR (Figure 2). He relaxed dramatically when seated in the Bead System (2).

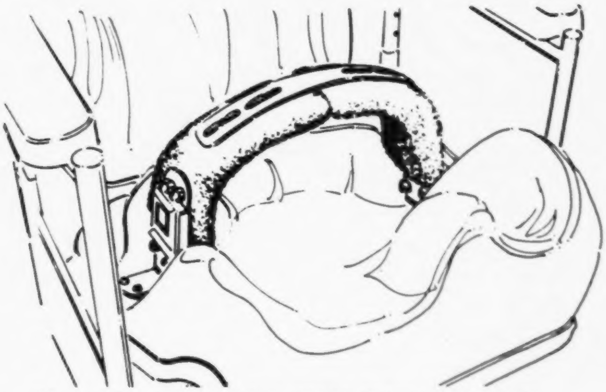

Figure 2 - New Modular Rigid Pelvic Restraint mounted on aggressively contoured Bead Seat.

The initial snug fit of the RPR prevented him from sliding out no more than 3/4" as measured from the front of his seat to the back of his calf. After this hour long trial Andy's skin was quite red an inch below his left ASIS. There was no such redness on his right side. It was evident that Andy sat with a subtle pelvic tilt. Loosening the RPR did not minimize the redness but only succeeded in allowing him to slide forward in the seat. After further observation it was obvious that the irritation was due to diaper snaps under the RPR. The RPR fit was tightened and new diapers were ordered for Andy. Further testing is required to see if the modular RPR will work as well as his initial version.

Sara C.

Sara a bright 14 year old athetoid C.P. lives at home and attends a progressive special school. Like Andy she is totally dependent for mobility and self care. Her Modular Plastic Insert System (MPI) (3) augmented by a chest panel, lap belt, and arm restraining tray (4), has failed to maintain her in a stable posture. Her athetoid movements and extensor thrust, while not as strong as Andy's, are sufficient to pull her down and out of the seating system. Efforts to tighten her lap belt only resulted in bruising her pelvis. In July 1984 Sara received a modular RPR to replace her MPI lap belt. Both Sara and her mother were very pleased with the way it kept her positioned in the chair.

In September 1984, Sara was having problems with the RPR fit. Her skin was red below both her right and left ASIS and she complained that the right side was painful. She had also slipped forward 1-1.5". The RPR fit more loosely due to the fact she was not wearing her occasionally worn diaper. Her strong ATNR was the cause of her asymmetrical pelvic discomfort.

In addition to tightening the RPR fit I moved it to a more distal location an inch below and forward of her ASIS. These modifications worked well until she began to wear heavier pants as the weather turned colder. In December 1984 her mother widened the RPR 1/4" and reported that this improved comfort while still holding her into the seat. Overall the RPR has been very well received by both Sara and her mother.

Mark T.

Mark, a 12 year old athetoid C.P., lives at home and attends a special school. He is also totally dependent for mobility and self care. He sits in a standard wheelchair which has been modified by the addition of a padded MPI seat and back. His athetoid movement patterns and ATNR to the right side are similar to those Sara demonstrated. Like both Sara and Andy a lap belt did not hold him back into the seating system.

In September 1984 I mounted a modular RPR snugly onto his padded MPI seat. This firmly locked his pelvis and he was unable to slide forward.

In October 1984 Mark's parents complained that even though they had raised the RPR 1/2" higher from the seat surface it was too tight - especially when he wore heavy blue jeans. Indeed it was very tight over his bony right ASIS. I loosened the fit making additional adjustments to eliminate the high pressures on the right side of his pelvis.

In spite of initial complaints of excessive tightness Mark and his family feel the RPR is a great improvement over his former lap belt.

Susan B.

Susan is a very intelligent 24 year old athetoid spastic C.P. quadraplegic. She lives at home with her family and is totally dependent for mobility and self care. She was referred to the UTREP-SP because her wheelchair insert was doing little to control her movement patterns and significant extensor thrust.

She received a Bead System wheelchair insert which, though better than her former system, was still not controlling her. In September 1984 UTREP-SP staff modified the Bead System by reclining it significantly. At the same time I added a modular RPR to the seat component. The combination of the reclined Bead System and RPR held her well.

In January 1985, Susan and her family reported that the modified seating/RPR combination was working very well.

Stevie, J.

Stevie is a 16 year old atheotid C.P. who lives at home and attends school three days a week. He too is totally dependent for self care and mobility. Physically he is quite similar to Andy - demonstrating random athetoid movements in combination with a powerful extensor thrust.

From 1976 - 1984 the UTREP service staff has provided Stevie with several seating systems in frustrated attempts to control his atheotid movements and resulting unstable sacral sitting posture.

In November 1984 Stevie received his most recent seating insert, a Bead System seat and back incorporating an RPR. The firm, intimate Bead System contours relaxed him considerably. The snugly mounted RPR successfully kept him back in the system. After sitting in the new system for two hours he had slid forward only 1/2".

Two weeks after receiving the system the RPR's fit was significantly loosened at the request of his family, who seldom use the device at home. His teacher reported that the loosely fit RPR fails to properly restrain him.

Stevie has not complained that the RPR hurt him, only that it was snug. After many years of marked sacral sitting, we feel he may not accept a more upright posture.

Lainey M.

Lainey is a bright 17 year old C.P. athetoid with a component of tension athetosis. She is dependent for self care. Aggressively followed by the UTREP-SP staff for her seating needs since 1976, she is well controlled by a Bead System with a lap belt, chest belt, and arm restraint tray. In January 1985 she received a new Bead System with RPR. She insisted that the RPR be snugly fit. Due to ATNR induced pelvic rotation and lower extremity asymmetry she quickly developed minor redness under her right ASIS and over her left proximal thigh. Initial feedback after the RPR adjustments to accommodate for her asymmetry has been encouraging. She most likes the positive RPR latching method which assures both her and her caregivers that she has been properly positioned.

DISCUSSION

Overall the RPR worked very well. For Andy, Sara, and Mark it provided enhanced pelvic stability which was unattainable by the use of a lap belt. For Andy and Sara it proved more comfortable than their former lap belts.

Andy's caregivers and Lainey herself liked the positive, fool proof latching method which insures proper positioning within the seating system. There were problems however which must be solved before the device can be easily and effectively fit on a clinical basis. A comfortable, snug fit proved very critical for the present RPR design. Adjustments of 1/8" - 1/4" caused significant changes in the device fit and function. RPR fit was significantly affected by changes in clothing bulk and subtle pelvic asymmetry.

CONCLUSIONS

A rigid pelvic restraint has proved more effective than the traditional lap belt in providing firm pelvic control for people with severe athetoid cerebral palsy. Further development of the concept is required before it becomes a readily usable clinical tool.

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P2.8

CUSTOM SEATING AT GILLETTE CHILDREN'S HOSPITAL

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ABSTRACT

Two major designs are used by seating services at Gillette Children's Hospital to meet the varied positioning needs of our clients. This paper discusses features and variations of these designs.

INTRODUCTION

Gillette Children's Hospital is a regional Health Care Center serving children and young adults with various disabilities. It is located in St. Paul and primarily serves people from the state of Minnesota, plus parts of Wisconsin, Iowa and North and South Dakota. Some people have traveled distances as far as from California and Alabama for services at Gillette.

Gillette Hospital provides services for all ages, although the vast majority are children and young adults. Their diagnoses range from cerebral palsy, myelomeningocele, muscular dystrophy, growth and developmental syndrome disorders, neurological diseases, to those who have sustained closed head injuries.

The client's living environments also vary from living at home with parents or foster parents, to group home facilities and state hospitals. Many of the people served require help solving seating problems that vary from very simple to extremely complex.

SEATING SERVICES AT GILLETTE

The designs we currently use have developed over the past 10 years and continue to evolve. We have found that optimum seating and positioning can be accomplished when a seating system meets certain basic goals. These include achieving optimum functional position and orthopedic alignment, positioning to inhibit abnormal reflex patterns, providing even pressure distribution and general comfort. It should also facilitate the work done by parents, caretakers or attendants. The seating system should enable the individual to enjoy increased function for daily living activities through compatibility with the living environment and transportation modes. This will enhance their physical and psycho-social development.

Evaluation System for making Seating Recommendations

The team approach is the key element of our Evaluation System. We include the client, his family, therapist(s), rehab. engineer, orthotist, adaptive equipment specialist, and medical staff. Information is gathered and discussed concerning the existing orthopedic status, functional level, present and future goals, the living environments (including home, work, school), recreation and transportation needs, and the needs of other users such as parents, teachers, friends, or attendants. Recommendations are made toward the specific type of seating system after the discussion is complete.

This team interacts not only at the evalution, but also on regularly scheduled follow-up appointments to assess growth and changes in the client's function and lifestyle.

TWO CUSTOM SEATING SYSTEMS AT GILLETTE

The Upholstered Sitting Support Orthosis

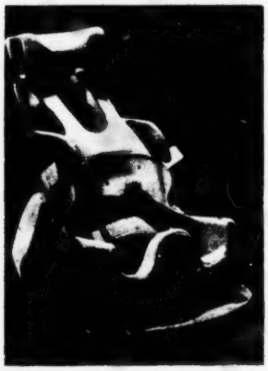
An Upholstered Sitting Support Orthosis consists of an ABS plastic shell that contains and unitizes the recommended combination of the following components: a firm seat and back, a pelvic belt, lateral pelvic bolsters, anti-thrust ledge, hip flexion wedge, pommel, lateral thoracic supports, lumbar spine support, anterior thoracic support (in the form of shoulder straps or a vest) and head support. It is important to note that each USSO is fabricated to the individual's specific needs and may be as simple as a firm seat, pelvic belt, and firm back; or as complex as to utilize all the possible components listed above.

Indications: Typically, a USSO is recommended for an individual that has minimal to fair sitting balance and has no significant spine or hip deformities. (It is not uncommon, however, to recommend a USSO for an individual with little or no sitting balance if there has been a spine fusion.) They usually have fair to good voluntary head control and have fair to good upper extremity function.

Features: The current USSO design is the product of a series of evolutions. Some of the design goals have been to achieve cosmetically acceptable appearances with unobtrusive borders and attractive naugahyde upholstering. They are portable and can be used in a wheelchair, on a couch, chair, or anywhere desired or needed. One useful feature is that they are light in weight, one-third less than the design using plywood, and are easier to handle. Comfort is assessed verbally, by the frequency of skin pressure problems, and by changes in sitting posture. Because most of our client population are children, the service life of each USSO varies by the growth patterns of each child. The average is 24-30 months in pre-adolescence, longer after adolescence. During the service life, the USSO is durable and is easy to adjust or modify to accommodate growth and/or lifestyle changes. Cleaning the USSO at home is easy in that all the pads and bolsters are attached to the ABS shell with velcro and can easily be removed. Another important feature is that components such as a headrest, pommel or anterior-thoracic support, can be made detachable to promote and facilitate ongoing therapy programs or for ease of transferring in and out of the USSO. We would like to acknowledge the many design ideas and features for the USSO which we have received from the Rehabilitation Engineering Center at Stanford and the Royal Ottawa Regional Rehabilitation Center.

Gillette type Sitting Support Orthosis

The Gillette type SSO is a custom contoured polypropylene shell mounted in a block of Ethafoam. We obtain an impression for the shell by positioning the client prone over an adjustable supporting fixture and utilizing the bean bag dilatancy casting method. This method allows us to maintain optimum orthopedic alignment, and also to accurately locate bony prominences for effective pressure distribution. The completed shell typically provides support from behind the knees to behind the head. The thin polypropylene allows bilateral thoracic support to extend as superior as the axilla. Components of the SSO include some combination of a pommel, pelvic belt, anti-thrust pad, pelvic growth adjustment pads, anterior thoracic support (either shoulder straps or a vest), flexible denim panels, shoulder protractor pads, special axillary supports, and a headrest.



Indications: An SSO is indicated for an individual with minimal or no sitting balance, and with fair to no voluntary head control. Typically, the clients are quite severely involved in their diagnoses, often rendering them more dependent and also more suceptible to debilitating orthopedic deformities. Various components can provide pressure distributions that are comfortable and can be strategically placed in an SSO to resist a deformtiy from occurring or progressing. If rigid orthopedic deformity is present, it is fairly simple to use this method to accomodate the deformity to provide support for sitting without excessive skin pressure.

Features: The ability to utilize aggressive positioning techniques is one of the primary features of the SSO. The SSO also shares many of the features of the USSO such as portability, light weight (from 0.7 kilograms to 4.75 kilograms), cosmesis, and comfort. Other features include the ability to employ unique components to help provide for special needs. These can be used to design flexibility into the system to allow for, as an example, periodic tone changes in some children with cerebral palsy. This includes removable components to promote or facilitate ongoing therapy programs or for ease of transferring in and out of the SSO. The service life of the SSO is presently 16-24 months. This, of course, can vary depending on the age of the child and the stage of their growth pattern. There are a number of growth adjustments that can be made to the SSO to accommodate growth and increase its service life.

SUMMARY

Typical optional accessories include a lapboard, custom positioning of the seating system in a wheelchair requiring modification to the wheelchair, and adapted joystick or other mobility controls.

Both seating systems can be designed to accommodate unique seating needs or to solve particularily unusual problems through the flexibility of a wide variety of applications. Gillette type SSO and USSO seating systems can and have been modified to enable children to use such recreational mobility devices as the "Gun It and Go" motorcycle and the "six wheeler all-terrain" cart.

Although the Gillette type SSO is designed primarily for use by individuals with cerebral palsy, we have found that with subtle design differences and the addition of a soft, non-elastic corset, it is an effective system for children with advanced Muscular Dystrophy. This combination provides aggressive support without loss of comfort and can enhance upper extremity function.

When it is practical, we combine commercially available components (such as lateral thoracic support pads, headrests, and seat cushions) with our seating systems, especially the USSO.

Although we are not providing commercially available seating systems through Gillette, we often recommend continued use of successful existing systems. Adjustments are made to existing systems, when appropriate, to maximize their effectiveness.

Since our clients are seen on site for various medical needs, follow-up to evaluate seating is easily coordinated with the client. This allows us the opportunity to closely monitor the fit and function of the seating systems, which is particularly significant because of growth and development factors in children.

Our experience has shown us that the wide variety of needs of each client cannot be met with one single approach to seating. The capability to use more than one design has enabled us to most effectively serve the wide spectrum of our clients' needs.

ACKNOWLEDGMENTS

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Please direct any questions regarding the seating program at Gillette to the authors at:

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FUNCTIONAL MOBILITY AND ADAPTIVE EQUIPMENT FOR LESCH-NYHAN CLIENTS

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ABSTRACT

This paper describes some relatively simple adaptive equipment that can be used to enhance the life of clients that are afflicted with Lesch-Nyhan syndrome.

INTRODUCTION

Lesch-Nyhan syndrome is a sex-linked congenital disorder. The disease is characterized by hyperurecemia and increased amounts of uric acid in the urine. The imbalances are caused by defective enzyme activity control. Stabilization of these chemical problems is accomplished through drug therapy.

Lesch-Nyhan syndrome is also characterized by choreoithetosis and behavior problems such as aggressiveness and self-mutilation. The self-mutilating behavior typically results in remarkable destruction and loss of tissue. Tissue loss around the lips and tongue, including partial tongue amputation, is most common. The mouth is used to mutilate the fingers, and the hands to mutilate head and face and other parts of the body. Self abuse also takes the form of attempted self-drowning, burning/scalding, mangling fingers in wheelchair spokes, and head banging. The drug therapy used to control the enzyme problems has no apparent effect on changing the behavior.

These children often display a tragic irony: Although their attempts at self-mutilation seem to be deliberate, most children when they acquire verbal ability, request full restraint at all times, to the point of suggesting that you hold their limbs more tightly upon transfer. While their physical actions of self-mutilation seem to be deliberate, they at the same time fear those actions. These children appear to be normal in their perception of pain.

Unfortunately, the behavioral aspects of Lesch-Nyhan syndrome usually lead to institutionalization of these children. From the onset of the disorder, parents and caretakers are unable to cradle, hug, touch and provide simple warmth and human contact. As these children mature, it becomes much more difficult to provide day-to-day needs. The physical exertion necessary for restraint during transfer from bed to chairs and for feeding exemplifies these problems. Without specialized adaptive equipment, it will take two or three people to safely transfer one of these teenagers.

To date few successful environments have been created wherein these children may functionally employ their limbs, independently mobilize, or learn via conventional methods which include hand function. Most frequently the child's hands and feet are tied to beds or wheelchairs to impose the immobilization these children demand and need. Helmets, mittens,and teeth guards re- used to limit the availability of hands to mouth. In some cases tooth extraction has been necessary. In Ottowa, a plexiglass shield was created under which a child's hands could be positioned on a table's surface. The plexiglass prevented oral mutilation of the phalanges, yet allowed some freedom for tactile/functional activities.

Gillette has had the opportunity to work with a child with Lesch-Nyhan syndrome over the past few years. A rather simple technique was developed to allow him to be more easily and safely transferred, to independently operate his manual wheelchair, and safely use his hands for drawing, reading, and operating a computer.

Client Description

The client is a fourteen year old boy who is currently living in a state hospital. He has spastic quadraparesis, choreoathetosis, and minimal mental retardation. His older brother manifests the same syndrome. He lived with his parents until it became too difficult for them to care for him. In foster care his self-destructive behavior required constant one-to-one interaction and supervision. He has attempted to drown himself, burn himself, mutilate his hands and feet in doors, and injure his head via repeated banging on chairs, walls, etc.. He has partially amputated two fingers as well as his tongue and lips. His behavior is disruptive because of occasional kicking, hitting and spitting at hospital staff. Despite these behaviors, he is a fairly attractive child with nearly normal communication abilities. He enjoys and seeks adult attention and carries on coherent and often enjoyable conversation. He wants to be with and "help" staff, yet is adamant about maintaining tight controls over his extremities so that he will not injure either himself or others.

When we began this project he was using a wheelchair in which he was positioned in a polypropylene sitting support orthosis. He wore a vest to prevent trunk flexion, a helmet, a tooth guard, and mittens. His hands were strapped to the anterior, distal portion of his wheelchair armrest or to his bed. Two to three persons were necessary to transfer him due to his strength and ability to coordinate hand to mouth movements.

Fabrication

Our initial goal was to facilitate safe transfer from bed to wheelchair. We fabricated a pair of mittens from pearled elk leather with one-half inch of high density upholstery foam. A draw string closure prevented easy removal. Since he could not bite through the gloves in a short period of time, one person could lift him and he could not damage his hands even if oral mutilation was attempted.

Our second goal involved independent mobility. Although he could sit untethered in his wheelchair wearing the mittens for short periods of time, the thickness of the mittens prevented him from using his hands to power his wheelchair. We needed to allow his hands the range necessary to wheel without allowing him the freedom to reach his mouth. While he was wearing the gloves, we attached a sheet of cardboard to each side of his wheelchair. The cardboard extended from the outside of the armrests to the wheel's axle, and also from the front to the back of the wheelchair side. We then traced the movements made by his wrists as we guided his hands through the motions required to wheel his chair and to move upward to the top of his lapboard.

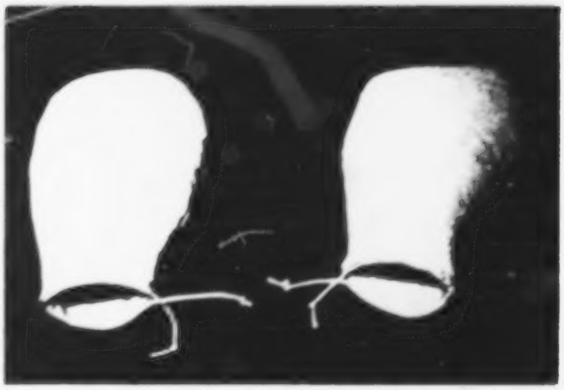
We then used the traced pattern to create a controued rod of 1/4" diameter mild steel. The contoured rod was attached to the modified, removable armrests of the wheelchair. Arm movement along the contoured rod was made possible by tethering wrist cuffs to the contoured rod by means of a "D" ring attached to each cuff and encircling the rod. The client's wrists could slide freely along the rod in the exact motion necessary for wheeling. They could also move to the top of the lapboard for function there. While his wrists were tethered to the contoured bar, the client's gloves, helmet, and vest could be removed without danger of mutilation.

Our third goal concerned improved use of hands for lapboard actitives. This goal was achieved by extending the contoured rod used to facilitate hands-on wheeling motion to facilitate hands-on lapboard activities.

Modified wheelchair used for Lesch-Nyhan client

Client positioned in chair with and without lapboard

Detail of contoured metal rod with wrist cuffs attached



Padded leather mittens used for transferring Lesch-Nyhan clients

Results

With these adaptations the client was able to move his wheelchair by himself. He no longer screamed for restraint when being moved and did not ask for his helmet while in the wheelchair. He now is helping to distribute supplies throughout the state hospital in which he lives. A wagon is attached to his wheelchair for his deliveries, of course someone else must load and unload his wagon for him. He is also interacting with a computer by means of switches mounted on his lapboard. We must note that none of this adaptive hardware was aimed at preventing our client from doing damage to his tongue and lips with his teeth. He, in fact, did have an episode recently where he further partially amputated his tongue.

Summary

The adaptive equipment described here provided increased freedom, mobility, and function for the client to explore his environment. It also provided a greater degree of freedom for those who are caring for him.

Note: Marcia Munson, formerly employed at Gillette Children's Hospital, is now employed at Fairview Hospital, Minneapolis, Minnesota.

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EARLY MOBILITY AID FOR NON-WALKING CHILDREN

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Mobility aid for children with multi limb deficiencies

Children who are born with multi amputations and limb deficiencies are severely limited in cognitive development because of the inability to function independently in an upright position. Piaget's theory states that object manipulation is necessary for successful early concept formation. Children with multi congenital amputations can develop early concept formation by observing the actions and words of people around them. Because the child's exposure to these activities is often limited by the amount of time that an adult is willing to donate, it is not surprising that some of these children have developed mental patterns not present in the "normal child".

Reviewing the literature, it was found that most children with these disabilities have been fitted, for mobility, with:

1. Bucket fitted to a caster cart (Fig. 1) (1).

2. Pylon-type swivel walker (Fig. 2) (2).

3. Electric cart (Fig. 3) (3).

All three of these options often present problems that cannot be overcome. The bucket on casters requires upper limbs or prostheses for propulsion, the swivel walker (which was very popular in Australia and Canada during the late 60's and early 70's but not widely used in the U.S.A.) requires great energy expenditures and as the child grows and the center of gravity is raised the walker may become dangerous, the electric cart has provided the best opportunity for mobility to this group but it is not widely used because each cart must be custom fabricated by a specialized center with most adjustments and repairs made at that center. The cost of returning to the center often is prohibited.

The Rehab Engineering Lab (REL) at the University of Texas Health Science Center at San Antonio recently was requested to help evaluate a two-year old congenital amputee with bilateral upper limb amelia, hip disarticulation on the right, and PFFD with hemimelia on the left.

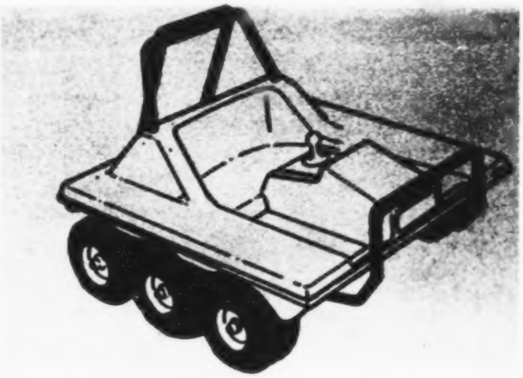
The child could move by log rolling, toys could be manipulated by head, neck and trunk movements, most toys had to be placed in position. The child could pick up objects with mouth and with the toes. The child seemed to have appropriate development skills: somewhat shy but very curious about her environment. The referring orthopedist was not in favor of prosthetic fittings at this time.

The rehabilitation team after the initial interview with the child and family decided that the best way to solve the immediate problems of mobility was to purchase an electric powered car sold in toy stores.

After viewing the available vehicles it was decided to buy a "Coleco" vehicle called G.I. Joe^{R} (Fig. 4). This toy vehicle is a battery powered three-wheel vehicle with a molded body. The 6-volt rechargeable sealed battery operates a motor drive assembly with two motors. The vehicle was modified by adding a ball bearing steering mechanism that the child could operate by holding it with her toes and pushing or pulling to make right and left turns. The motor wiring was modified to accept a microswitch that would come on when the steering mechanism was lifted. It would automatically turn off when the steering mechanism was released. A second microswitch that put the vehicle in a reverse mode was attached.

The child came to the REL for an introduction to the modified vehicle. A remote controlled switching arrangement was adapted so that the vehicle could be moved without sitting in it. The child was able to make the vehicle move while sitting in the mother's lap. After a few minutes the child was placed in the vehicle and soon was operating the forward and reverse switchings.

The G.I. Joe^{R} proved to be an easily modified "off-the-shelf" vehicle that cost only a small fraction of a specially designed and fabricated electric cart, its only drawback was the installation of a steering mechanism. The Rehab Team wanted a system that could be put in place with no professional help (such as the family) and at a minimal cost. It was decided that a "joystick" system for total operation of the vehicle was the most appropriate way to proceed. Plans were drawn up to adapt a low cost video game joystick to the G.I. Joe^{R}. Before these plans were finalized a new electric powered toy vehicle, "The Probe VI", (Fig. 5) made by Hedstrom Corporation was located. This vehicle comes complete with two 6-volt gel batteries, two 6-volt electric motors, it is a chain drive joystick operated vehicle with rubber bumpers and a tow-bar.



The joystick allows forward, reverse, right and left turns, that can be made in a very short radius, because there is a chain drive on each side with its own motor.

The child was presented with the new vehicle (Fig. 6) (the only modification was a safety strap and foam padding to aid balance) and was immediately able to operate the joystick in the forward and right turn modes.

The family was instructed in the vehicle's operation and sent home to practice. The total cost of the cart was $170.00. This vehicle should be sufficient as a mobility aid for several years with proper training and supervision a vehicle of this type can be used as a mobility aid for children aged 18 months to 7 years.

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*Photography done by Cono Farias, Photographic Technician II, Radiology Department, University of Texas Health Science Center at San Antonio.

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TWO MOBILITY DEVICES FOR CHILDREN WITH SPINA BIFIDA

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ABSTRACT

Mobility provides young children with the opportunity for many developmental experiences. Children with spina bifida have physical limitations which hinder these developmental experiences. This paper describes two assistive devices designed to provide age-appropriate independent mobility, the Caster Cart and Crutchless Standing Orthosis.

INTRODUCTION

Children with physical limitations are denied many normal experiences, usually taken for granted, that enhance social and psychological development. A wide variety of assistive devices are available to provide some of these experiences in an age-appropriate manner. This poster presents two devices developed and fabricated at Gillette Children's Hospital in St. Paul, Minnesota.

A large number of the clients served at Gillette are children and young adults with Myelomeningocele or Spina Bifida. The majority of these young children have significantly limited muscle function of their lower extremities. Normal children are becoming very mobile on the floor before one year of age, yet this mobility is physically difficult or impossible for infants with Spina Bifida. The devices presented here provide important mobility and hands- free standing as a means to more normal, enjoyable and useful activities for the child.

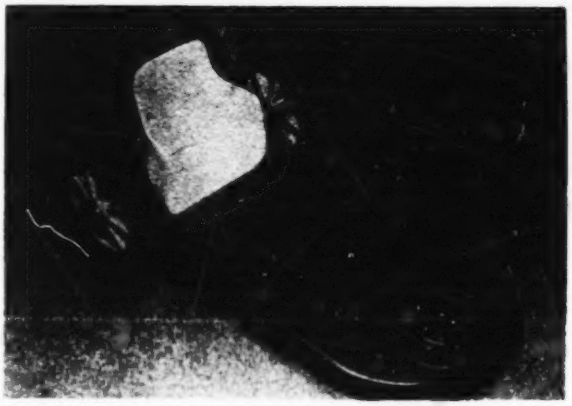
Preliminary designs of these devices were adapted in the early 1970's from several used in a child paraplegic program developed by Wally Motlock and others at the Ontario Crippled Children's Center. The devices currently fabricated at Gillette have several differences, described below in Design Features, that have increased their usefulness for our clients.

CASTER CART

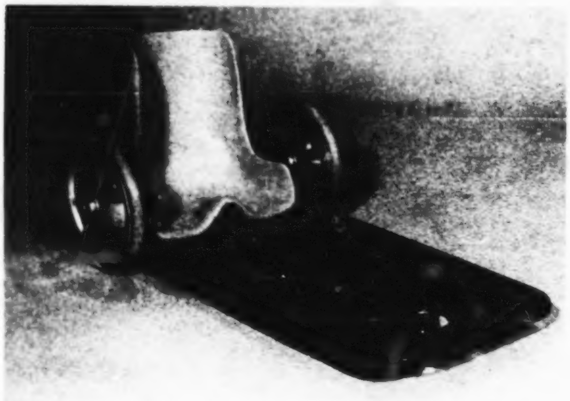
Discussion of Need

Young children use a remarkable amount of mobility in discovering their environment and building their self-image. Limitations to a child's independent mobility naturally decreases opportunity for many early developmental experiences, such as exploring one's environment, discovering cause and effect relationships, or interaction with one's peers. The caster cart fabricated at Gillette, described below, is simple in its design, but has been used to provide independent mobility to a large number of our young clients with Spina Bifida.

The very close proximity to the floor (about 1-1/2") allows independent transferring by the child, and it's use simulates sitting and playing on the floor. There is also extra room on the cart to carry toys and pets while keeping hands free to propel the cart's wheels. The caster cart's design allows free rolling and is highly maneuverable. Sitting support is provided as needed.



ABS Plastic Shell Design



Aluminum Shell Design

Design Features

a. The caster cart is positioned 1-1/2 inches above the floor for independent access and peer-level interaction.

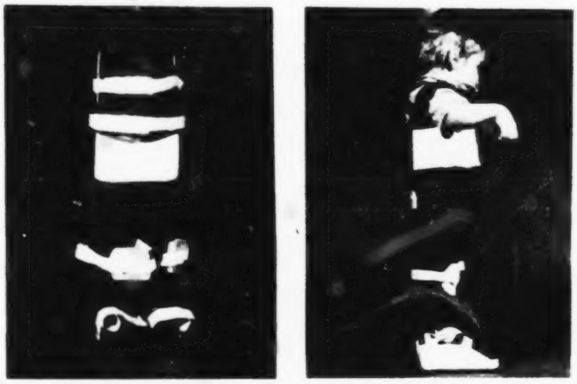
b. We have used to different shell designs, both light in weight. Design drawings for an aluminum shell are available and may be easier to fabricate in certain facilities.The shell of the cart we currently provide is vacuum-formed from ABS plastic. The contoured edge and two base-mounted aluminum bars contribute to the shell's rigidity.

c. Sitting support is provided by a contoured seat, as shown, which is made of polypropylene. Minimal support is supplied to give required trunk stabilization with maximum upper body freedom.

d. There are two large wheels and one caster attached to the caster cart. The two large wheels are located on each side for propulsion and the caster is located beneath the front center portion of the cart. Aluminum plate inserts are vacuum-formed into the sides of the shell to strengthen the axle mounting. An important feature of this cart is the high quality wheels, which maximize the cart's maneuverability and minimize the energy required for propulsion.

The wheel design differs from the O.C.C.C. cart in that the caster is smaller and located forward. The side wheels are located more toward the back of the cart which gives better lateral access to the floor.

CRUTCHLESS STANDING ORTHOSIS (CSO)



Discussion of Need

Standing is a position that has physical, developmental, and psychosocial benefits. The Gillette type Crutchless Standing Orthosis (CSO) has developed over the past 11 years and is presently custom fabricated for many of our Spina Bifida clients from ages of one year to six years. Some users choose to continue use of their CSO up to age nine or ten. The CSO is designed to accommodate growth with simple modifications. Donning and doffing is easily and quickly accomplished which helps the CSO be well accepted by both children and parents.

The minimal weight of the orthosis is very important, particularly for use with the small child. Gillette recognizes the quality of other standing orthosis designs but we have chosen our own design to decrease weight.

There are several physical benefits of the CSO. Daily weight-bearing sessions on lower limbs will maintain bone strength and decrease tendency toward osteoporosis and recurrent fractures. The upright position is reported to also improve over-all urologic function in addition to general improvement in circulation and respiration. Standing will also provide additional stretch to assist non-progression of muscle contractures, particularly about the hip and knee.

The social benefits of the standing experience can be dramatic. As peers begin to walk, effective interaction is enhanced when eye-level positioning is provided. Interaction is also enhanced when mobility is provided to move freely and play with peers. With a small amount of upper limb strength, the child can soon begin to move about with a walker. Many of our children with good balance and coordination learn to "swivel walk", with their hands free, through a rhythmic thoracic rotation. Upright positioning and working toward locomotion will exercise residual muscle strength. Playing ball, pushing a baby buggy, brushing teeth and helping with household chores are some examples of activities of our clients using a CSO.

After entering grade school, children continue use of the CSO. Larger physiques and decreased energy levels can make the use of the CSO impractical after this age. If daily routine indicates the need for frequent sitting, a child may choose to discontinue use of the CSO and consider replacing it with the heavier, jointed parapodium. It has been our experience that children naturally come to these decisions.

Since many children with spina bifida require spinal orthotic treatment, it is important to note that the CSO is entirely compatible with any spinal orthosis.

Design Features

a. There are three points of pressure applied to extend the hips as shown. 1. Abdomino-thoracic apron, 2. plate behind buttocks, and 3. well-padded knee supports
b. The CSO is light in weight and averages 1.24 kilograms, depending on the weight of the child. The frame is fabricated of 1/2" single anodized aluminum tubing, alloy 6061-T6. Modifications to height during fit and especially during growth are easily made by adding a minimum of telescoping 5/8" tubing. An anodized aluminum plate is located at the buttocks to provide frame structure and postural support. Anti-tip bars are added to increase stabilization against backward falls.
c. The foot sockets are vacuum-formed of ABS plastic and attached to the aluminum frame's base. Leg length discrepancies are easily accommodated by the addition of dense foam spacers to the bottom of the ABS foot socket. The bottoms of these sockets are covered in leather to reduce floor friction which facilitates easier locomotion. The anterior lateral edge of the leather base is slightly reduced in thickness to also facilite swivel-action motion
d. Trunk freedom is provided whenever possible for maximum locomotion ability. Lateral bolsters, however, are mounted to the CSO frame when indicated.

Concluding Statements

The mobility devices described in this paper have been used widely by our client population. Some mothers feel the caster cart has been the single most useful tool for their child.

Efforts are being made to explore the feasibility of commercial production of these devices. Custom-fabrication can easily be accomplished for small quantity applications and specification drawings are available at Gillette.

Technical support for the fabrication of these devices has been provided by David Lawson, Rick Weber and others of the Orthotic and Prosthetic Lab at Gillette. Please direct any questions regarding the fabrication and usefulness of these mobility devices to the authors at:

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ACCESS TO OUTDOOR RECREATION: TWO ADAPTED GEAR DESIGNS

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ABSTRACT

This paper describes solutions to some of the difficulties encountered by mobility impaired persons participating in outdoor wilderness experiences. The devices described are the People Pack and modified Pulk snow sled.

INTRODUCTION

It is increasingly common to hear of able and disabled groups of people enjoying the same recreational and outdoor activities together. One organization established in the Twin Cities, Wilderness Inquiry II, plans and leads many vigorous trips lasting from 5 to 28 days. They organize groups of able and disabled participants on such summer adventures as canoe trips, backpack hiking and camping, and such winter adventures as cross-country skiing, snow-shoeing, dog sledding and pulk sledding.

Gillette Children's Hospital began working with this organization when they were plagued by two persistant problems. The first problem was assisting those disabled participants with limited upper body strength across rough terrain, such as trails or portages, during hiking and canoeing activities. The other problem was the instability of a pulk sled used by disabled persons with similar physical limitations for cross-country and snowshoeing expeditions.

The Orthotics and Prosthetics Lab at Gillette has worked closely with members of this organization to design and field test solutions to both these problems. Through a minimum of adapted equipment, these problems have been largely eliminated by (a) the People Pack, a system by which a non-disabled person can easily carry a disabled person and (b) modifications to a Pulk sled which significantly increases its stability.

PEOPLE PACK

Problem Discussion

In crossing short areas of rough terrain, such as portages, during canoe trips with groups of people of various disabilities, it is highly efficient for non-disabled persons to assist in the transportation of mobility-impaired persons. Persons with physical limitations in their lower extremities, only, are able to assist in the transportation by holding tightly to a carrier person in a piggy-back style. Persons with physical limitations in both upper and lower extremities present a challenging problem for their easy transportation.

The design of an assistive transportation device must be light in weight and should be streamlined due to the need to conserve weight and space in packing. The device must comfortably suspend the disabled user and comfortably distribute weight across the shoulders and back of the carrier.

Prototype I

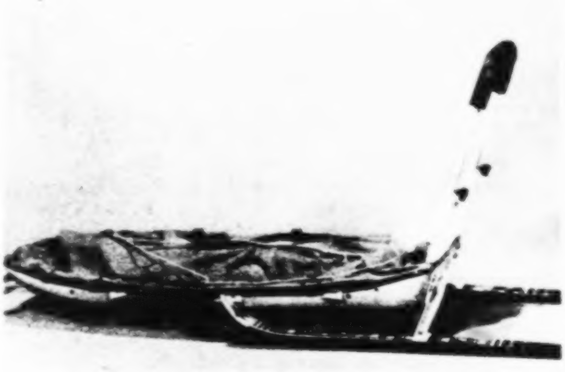
Design: The original prototype of the People-Pack was fabricated from a commercial tubular aluminum back pack frame. This frame was chosen due to its proven ability to distribute great loads over the shoulders and back of users. A custom-made canvas support sling was designed to support the entire weight of the person being carried. This sling holds the person beneath the buttocks and thighs, and provides sitting support laterally and posteriorly across the back. After the sling is positioned around the person, it is then connected to the back pack frame by two wooden dowels which insert into the top of the frame. The mobility-impaired person is lifted into position on the carrier's back by one or two assistants.

Results: Although this prototype was a great improvement over no assistive device, through field testing we found that the back pack frame itself was uncomfortable to the person being carried. The width of the frame impinged on the medial aspect of the thighs and across the sternum. The canvas sling was somewhat difficult to don and it also was not considered cosmetically appealing.

Prototype II

Design Improvements and Results: The second design, after one year of field use, seems to have solved the problems encountered with the first prototype. A specially designed frame was constructed out of 1-1/2" x 3/16" aluminum bar stock and contoured with a lumbar roll. This frame is more comfortable to the carrier because of the contouring and the large flat surface contact. The contoured design distributes the load across the back more evenly than the commercial flat surface contact. We attribute this result to the difference in load bulk The narrow 8" frame width and the streamline contour to the carrier also eliminates any impingment on the person being carried. The canvas support sling was replaced with oversized denim jeans modified by shortening the legs to knee length, reinforcing the thigh area and adding suspension straps. The "shorts", by comparison to the canvas sling, are easy to don, are cosmetically acceptable and still capitalize on weight-bearing through the thighs. The harness system utilizes rectangular loops that slide onto frame extensions, as shown in the photo. This attachment method is much easier to use than the former wood dowel peg system. Padded straps, attached to the back of the "shorts" and crossed over the back of the person being carried, provide adequate trunk support.

PULK SLED



Problem Discussion The fiberglass pulk sled was originally designed to be a cargo sled with a typical low center of gravity, and it works well for that application. It becomes quite unstable, however, when it is used to carry a person over rough or uneven terrain. For this application the center of gravity is significantly heightened and is disproportionately located over the rear of the sled. Instability has not been a serious problem for riders with upper body strength since they are able to provide occasional support when the sled begins to tip sideways. It has been a very difficult and annoying problem, though, for riders with upper body involvement. These riders also lack sitting balance and found themselves sinking into the sled when no sitting support is provided. In addition to the sled instability and postural problems, the shallow tracks on the pulk sled caused the sled bottom to drag which increases friction and required unnecessary energy expenditure on the part of the person pulling the pulk sled.

Design Modification: To solve the instability problem, we first widened the base of support with outriggers of shortened downhill skis. These were mounted to the rear of the pulk sled over a 1/4" aluminum plate. This plate was used to strengthen the fiberglass sled in this area since high forces will be applied when the outriggers provided support. A matrix of mounting holes on this aluminum plate allow for on-the-trail adjustment for a variety of snow depth conditions. Trunk support was provided by a telescoping aluminum frame, canvas back panel, and an elastic thoracic strap.

The track depth on the bottom of the pulk sled was increased by 3/4" using maple board shims beneath the 3/16" polypropylene tracks.

Results: This prototype was tested for one season with great success in avoiding tipping due to instability. Pulling resistance did not increase because of the outrigger skis and was, in fact, improved by the combination of greater sled stability and increased bottom track depth. A manufacturing company has since produced 6-8 models from this prototype.

ACKNOWLEDGEMENTS

The authors would like to acknowledge the wide variety of people involved in these projects. They include Greg Lais and many others from Wilderness Inquiry II, and J. Martin Carlson, Director of the Orthotics and Prosthetics Lab at Gillette Children's Hospital.

We hope these adapted gear designs can benefit other adventurous explorers. Please direct any further questions on design or applications to the authors at the following address:

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THE SUNBURST TANDEM

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Palo Alto Veterans Administration Medical Center

ABSTRACT

The Human-Machine Integration Group within the Palo Alto VA Rehabilitation R&D Center has endeavored to develop devices responsive to the recreational, mobility and therapeutic needs of individuals with disability. One such project has been the design of a tandem bicycle for able-bodied and disabled individuals to ride together, called the **Sunburst**. Individuals with various physical abilities, resulting from paraplegia, low level and incomplete quadriplegia, amputation, muscular dystrophy, multiple sclerosis, cerebral palsy, blindness, etc., may now enjoy the physical, emotional and social benefits of riding a bicycle. As a product intended to appeal to able-bodied individuals as well, the enhanced marketability of the **Sunburst** should lead to a less expensive quality product with greater availability.

INTRODUCTION

Bicycling is a revitalizing activity, whether for recreation, sport or self-powered transportation. The physical exertion, the sensations of motion and balance, and the opportunity to share a popular activity with friends provide the bicyclist with fitness, a clear mind and a renewal of spirit and enthusiasm that carries into other activities of daily living.

With the availability of the **Sunburst** tandem, individuals with various disabilities will be able to share in the freedom, exhilaration and accomplishment of riding a bicycle.

FUNCTIONAL DESCRIPTION

As a natural spinoff from the **Handbike** (1,2,3,4,5), an arm-powered bicycle developed at the Palo Alto VA Rehabilitation Research and Development (RR&D) Center, the **Sunburst** tandem combines arm- and foot-powered recumbent cycling in the front with a standard bicycle in the back (Figures 1, 2 and 3). The back rider steers through a remote linkage from the handlebars to the front wheel, sitting high enough to comfortably see over the front rider. With this configuration, both riders have a clear view ahead.

The back rider controls the rear caliper brake with a hand lever. In addition, both riders activate the front hub brake by backpedaling.

The front rider pedals with any combination of arms or legs. The front cranks directly couple allowing the front rider to assist or passively exercise his or her own less functional limbs. Alternatively, a leg rest attaches to the seat (Figure 2). A freewheeling system allows the front rider to stop pedaling to rest while the back rider continues to pedal. The **Sunburst** riders move as one, combining their pedaling effort and balance to their best ability.

When the **Sunburst** comes to a stop, the back rider is in a good position to hold the bicycle upright, as foot placement on the ground is approximately midway along the length of the bicycle. The front rider then lowers a kickstand which latches at each side of the seat.

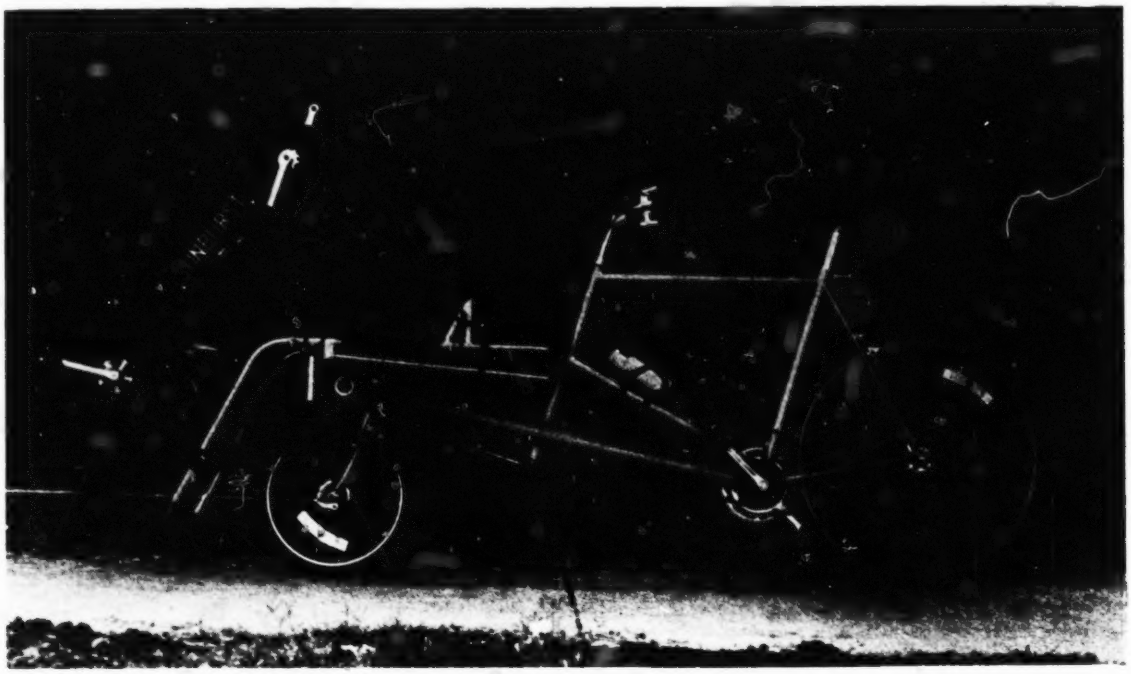

Figure 1. **Sunburst** tandem.

Figure 2. **Sunburst** tandem showing use of leg rest.

Figure 3. **Sunburst** tandem with able-bodied riders.

CONSTRUCTION

The Sunburst uses many standard bicycle components and custom frame building technology. High strength chromium molybdenum steel tubing is mitered and the joints are brazed without lugs. This allows flexibility in configuration and a relatively quick turnaround in prototype iteration to incorporate the latest design improvements in each production run.

PERFORMANCE

Initial informal evaluation of the **Sunburst** has begun to show its potential. Further test riding and evaluation should demonstrate that the flexibility of the front drive system in a recumbent tandem configuration will accommodate various physical abilities resulting from paraplegia, low level or incomplete quadriplegia, amputation, multiple sclerosis, muscular dystrophy, cerebral palsy, etc. Also, because the back rider steers, the front position is also well suited for a visually impaired rider. With the front position designed to appeal to able-bodied riders as well, an individual with disability has the dignity of riding the **Sunburst** in the same position as his or her able-bodied friends.

AVAILABILITY

The **Counterpoint**^{TM} is another tandem similar in configuration and developed independent of the RR&D Center's **Sunburst**, by Jim Weaver of Counterpoint Conveyance Ltd. (P.O. Box 33475, Seattle, Washington 98133, 206/365-6837). The **Counterpoint**^{TM}, which has received patent approval, was first offered with foot power only, and targeted an able-bodied market. Later, an arm crank was added to either side of the front rider on a **Counterpoint**^{TM} built for an individual with lower limb paralysis.

Starting with a different product orientation, the first tandem prototype developed at the RR&D Center, called the **Handbike Tandem**, featured only arm-power for the front rider, and was intended for use by individuals with disability (1). The **Sunburst** is the second tandem prototype developed at the RR&D Center, and includes arm- and foot-power in the front as described above.

The author and Mr. Weaver became aware of each other's designs and began to exchange information. Since then, Mr. Weaver has experimented with arm cranks directly in front of the front rider (for a recent customer with paraplegia), and with arm and foot cranks combined, similar to the **Sunburst** - both configurations are now available. In short, the **Sunburst** concept is essentially now on the market as one version of the **Counterpoint**^{TM}. Further collaboration is being explored to combine the best of both designs and evaluate.

ACKNOWLEDGEMENTS

The **Sunburst** and its earlier prototype, the **Handbike Tandem**, have been developed at the Palo Alto VA Rehabilitation Research and Development Center. The **Handbike Tandem** project was carried out in conjunction with the University of British Columbia, Department of Athletics, under a grant from the British Columbia program for the 1981 International Year of the Disabled Person. Custom frame building and design consultation were provided by Gary Hale of Gary Hale Bicycles (1156 W. Second St., Eugene, Oregon 97402, 503/342-4583).

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P2.14

BASIL PROJECT - ECR III

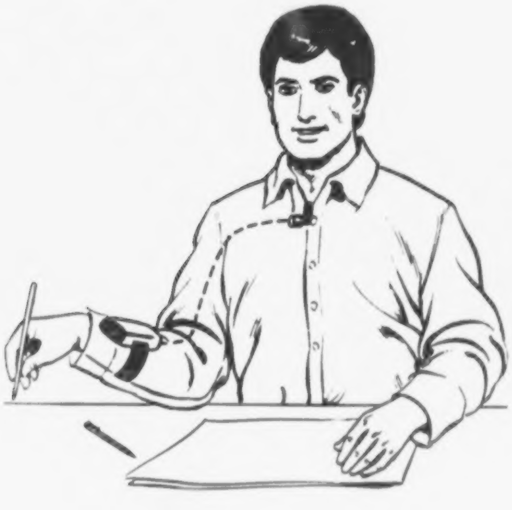
JAIME OCTÁVIO DE MAGALHÃES FILIPE
APC-CIDEF-CENTRO DE INOVAÇÃO PARA DEFICIENTES-PORTUGAL

BASIL PROJECT

The CIDEF - (Innovation Centre for the Handicapped) -, through its Scientific Research Department, and within the scope of Rehabilitation Engineering, has been developing the prototype of a device intended to give a "certain type" of information on the presence of sounds to people who are completely deaf. This project has been named BASIL.

BASIL is a device intended to "inform" a deaf person of the proximity of sounds such as "the crying of a baby", "the ringing of a telephone", "the buzzing of a door-bell", or the proximity of the noise of a car engine, a car horn, etc.

It is a small, unsophisticated, device, consisting of the following main items: a mini lapel-microphone of the Electret type, a monobloc box containing a binary expanding power amplifier, a small electric motor and an accumulator.



By means of this device, any medium or loud sounds caught by the microphone will be amplified and will cause the rotation of a motor with which an eccentric accessory is coupled, causing it to start a strong vibration. The vibration produced by this small motor will be transmitted to the monobloc bow which in turn will transmit them to the body of the handicapped person by contact with the skin.

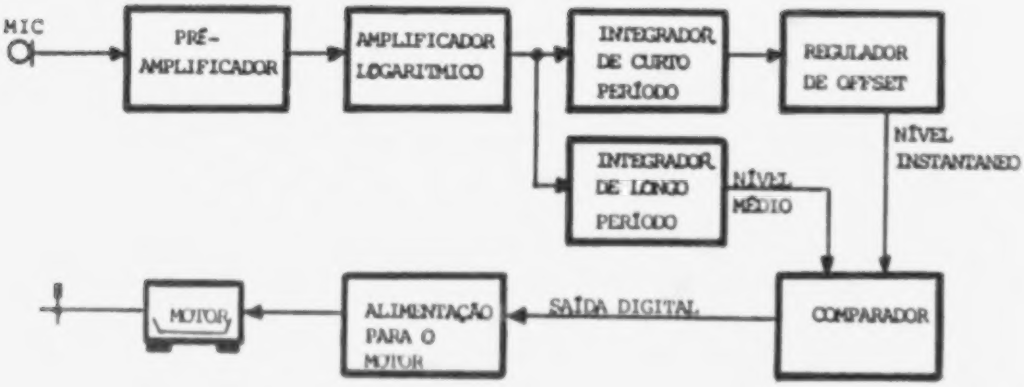
Two different sound levels have been taken into consideration: the medium level and the instantaneous level. The medium level corresponds to the ambient sounds and does not activate the device. The instantaneous level, having a higher intensity, will cause the motor to start rotating. However, if the instantaneous level is prolonged, then it will be interpreted by the binary system as a medium level and the motor will stop.

It is possible nevertheless to vary the sensitivity of the device, so that BASIL may supply indications relating even to distant and low-level sounds.

The device may be attached to the waist or placed on the arm in which case it should be near the wrist and held in place by means of a bracelet.

This project has been created by the writer, and its electronic conception is the work of Pedro DE SOUSA, a 19-year-old Engineering student, assisted by another Engineering student, 20-year-old Carlos GRAVINHAS.

It is hoped it will soon be industrialised through the Portuguese Electronics Company "CENTREL".


MIC
PRÉ-AMPLIFICADOR
AMPLIFICADOR LOGARITMICO
INTEGRADOR DE CURTO PERÍODO
REGULADOR DE OFFSET
NÍVEL INSTANTANEO
INTEGRADOR DE LONGO PERÍODO
NÍVEL MÉDIO
COMPARADOR
SAÍDA DIGITAL
ALIMENTAÇÃO PARA O MOTOR
MOTOR

BASIL - ESQUEMA BLOCO

ECR III

Another project which is being developed by the CIDEF is a system for the manual lifting of wheel-chairs from street level to the inside of vehicles, such as vans.

This is a device consisting of two rails inside which there run chains of the mini-bicyle type, and which by means of a handle will lift a platform carrying the handicapped person in his wheel-chair. With little effort it is possible to lift a weight of around 100 Kgs. up to the level of the van floor.

The ECR III can either be disassembled and unfolded, so it can easily fit into a van, or be permanently set up in order to ready for use by simple rotation.

Project created by the writer.

THE POSTURE SAFETY SEAT

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ABSTRACT

This paper describes a handicapped car seating system called the Canadian Posture and Seating Centre (CPSC) Posture Safety Seat. Handicapped adults require a seat insert and restraint design that provides physiological support as well as safety. The Posture Safety Seat incorporates the tried and proven concept of modular seating for support and has undergone several research crash tests towards the development of a safe restraint system.

INTRODUCTION

Presently car seat carriers are in existence for occupants up to 48 pounds. Canadian Motor Vehicle Safety Standards (CMVSS) are quite explicit as to their design requirements. For occupants over 48 pounds however, no explicit standards or approved system exists that provides the same kind of safety and support as the children's car seats.

The purpose of this project has been to develop a car seat insert to meet the needs of the handicapped population over 48 pounds. The project proposes the utilization of a modified CPSC Winnipeg Modular Seating System in conjunction with an additional restraint system integrated with the automobile's lap belt in the rear seat. Parallel research results are forthcoming from Transpoprt Canada which will eventually determine CMVSS standards for car inserts for handicapped occupants.

BACKGROUND

The CPSC Winnipeg Modular Seat consists of a standard back and pommel seat pan. The seat and back modules are vacuum-formed ABS plastic shells which are upholstered with vinyl covered foam cushions that are velcro attached. The standard back has integral side bolsters for lateral support and the standard pommel seat is designed with a 15 degree anterior wedge and a centre pommel. Figure 1 shows the critical dimensions of the standard back and pommel seat. To form the system the back is bolted to the seat pan thereby maintaining the back to seat positioning.

The Modular Seat system was designed to provide special seating at minimal cost for clients of various sizes and disabilities. The mild to moderately handicapped form the predominate population of users.

The original idea of an adult handicapped car seat was to provide a kit that could adapt any Winnipeg Modular Seat for safe automobile use. In order to test the plausibility of this idea an investigative crash test was arranged at the Impact Facilities at the Defence Civil Institute of Environmental Medicine in Downsview, Ontario.

From the test many modifications and improvements were made on the design. Three addition research crash tests have been performed since, and from which general observations indicate favourable results.

THE DESIGN

The posture Safety Seat is comprised of two main components the posture seat and an add on safety kit. The posture seat is simply a standard CPSC Winnipeg Modular Seating system and the add on safety kit consists of a harness, submarine belt, tummy pad and side plate modifications. The posture seat with the safety kit installed is shown in figure 2.

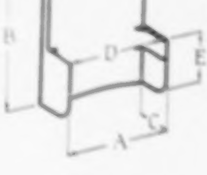
The Safety Kit

The first safety kit was simply a shoulder harness anchored behind the occupants head and then looped down around the car's lap belt. An investigative crash test on the original system indicated several problems which were:

1. The shoulder straps pinched inwards on the occupant's neck.
2. The lap belt rode up over the illeac crests of the occupant.
3. The lap belt twisted into a rope.
4. The seat insert shifted forward on the car seat bench.

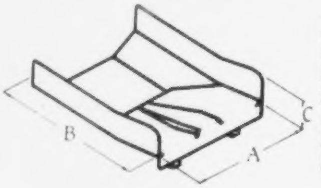
FIGURE 1

CPSC Winnipeg Standard Back



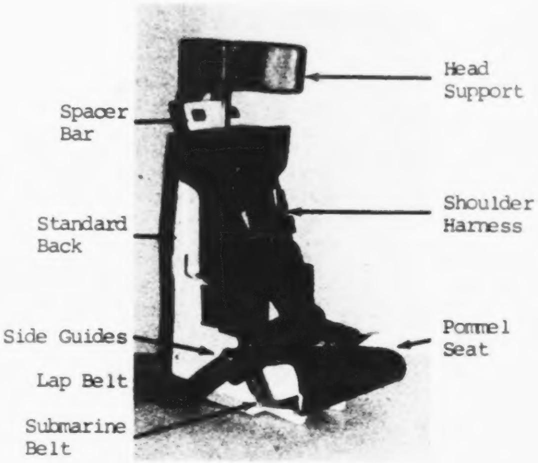
Product#	A	B	C	D	E
SB-01	10.5	19	3	7.75	6
SB-02	11.5	19	3	8.75	7
SB-03	12.5	21	4	9.75	8
SB-04	13.5	22	4	10.75	8
SB-05	14.5	23	5	11.75	9
SB-06	15.5	24	4	12.75	9

CPSC Winnipeg Pommel Seat



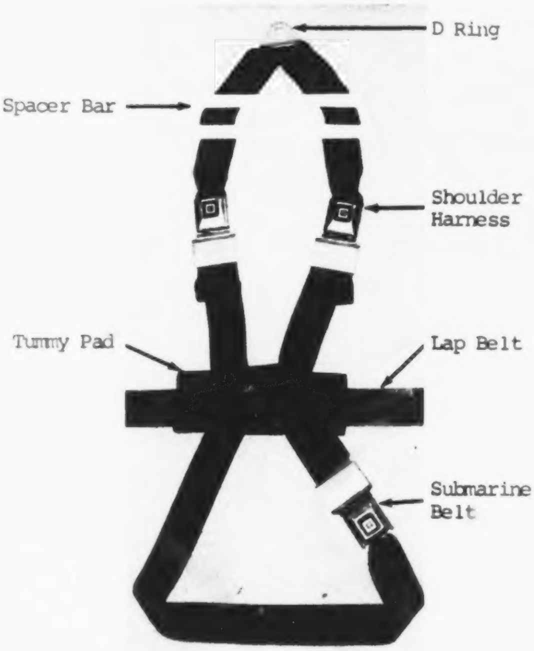
Product#	A	B	C
SP-01	11	12.5	4
SP-02	12	13.5	4
SP-03	13	15	4
SP-04	14	16.5	4.5
SP-05	15	18	4.5
SP-06	16	20	5

FIGURE 2
The Posture Safety Seat



In order to prevent pinching of the occupant's neck and maintain the shoulder straps on the clavicle area a spacer bar has been integrated into the harness either behind the occupant's head or on their chest. The spacer bar behind the head is shown in figure 3. The chest spacer is similar except that it is rectangular in shape. Available space in the automobile's rear window deck will determine which type of spacer should be used.

FIGURE 3
The Safety Kit



The next problem encountered was the 'riding up' of the lap belt. This is largely due to the upward load exerted by the shoulder harness during impact. Children's car seats make use of a crotch strap which works quite well so a similar type of belt was incorporated into the Posture Safety Seat design. The submarine belt (see figure 3) loops around the seat pan and over the lap belt to prevent 'riding up'; the underside structure of the pommel seat prevents the belt from sliding.

Thirdly the 'roping' of the car's lap belt was alleviated by using a tummy pad with a stiff internal structure. The tummy pad shown in figure 3 is constructed of 1/8 inch thick ABS plastic covered with naugahyde. The plastic insert is slightly curved to accommodate the occupant's shape and the pad constructed to allow the lap belt to thread through it. The shoulder strap then loops under the lap belt and the submarine belt loops over the lap belt.

Finally the shifting of the insert was prevented by side hooks or guides for the lap belt (see figure 2). Similar guides are used on car booster cushions. For the Posture Safety Seat, a side template is glued to the sides of the pommel seat pan and a circular cut out is made to form the lap belt guide. This guide maintains the position of the lap belt on the insert thereby holding the insert in place.

The Posture Seat

The CPSC Winnipeg Standard Back and Pommel Seat are made of tough 1/4 inch ABS plastic. No fundamental design changes of the plastic shells are required as yet, except for the minor side modifications on the pommel seat pan. The one point which needs to be considered for safety is the head support. Present CMVSS require that headrests undergo either a dynamic or a static test. The particulars of the tests can be found in Section 202 of the CMVSS. Before testing, calculations have been done to form a design comparison. This comparison was made between an ordinary Ford car headrest and the CPSC head support. The assumption made in these calculations is that the headrests could be simulated as cantilevered beams.

	Ford	CPSC
Dim.	0.25"*1.5"	0.5"*0.5" (0.0625" wall)
Mat'l	steel	steel
Const.	flat stock	square tubing
Length	max. 8"	max. 8"
D.L.L.	68 lbs	62 lbs
P.L.L.	102 lbs	98 lbs

Note: D.L.L.= Deformation Load Limit
P.L.L.= Plastic Load Limit

Since the Ford headrest passes the CMVSS tests, this comparison would indicate that the CPSC head support would pass the same tests.

DISCUSSION

From the research crash tests, before and after analysis indicates satisfactory performance but a frame by frame analysis of the high speed films is required. This analysis will provide quantitative data on the dynamic performance of the Posture Safety Seat. Concurrently the standards being developed by Transport Canada from their research will have an impact on the design. The film analysis and the Transport Canada research will determine the final design suitable to begin compliance testing.

The vacuum formed CPSC Winnipeg Modular Seating incorporated into the Posture Safety Seat provides the extra support and postural positioning required by the clients. In addition the quality reproducibility of vacuum forming is approximately 98 per cent. This means a consistent quality product.

The present safety kit would have retrofit capabilities for present owners to adapt their Winnipeg Modular System into a Posture Safety Seat. Also, the harness would be manufactured by an approved seat belt manufacturer because they have the facilities to ensure a safe and compliance standard restraint system.

In so far as research and development have gone the CPSC Posture Safety Seat represents a safe viable solution to the problem of providing a car seat insert for handicapped adults. Once Transport Canada's research is complete, the design can be finalized and compliance testing to the CMVSS standards can begin.

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IMPROVED MYOELECTRIC PROSTHESIS CONTROL USING MICROPROCESSORS

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ABSTRACT

The design, implementation, and testing of a microcomputer based myoelectric limb controller is described. Its advantages and disadvantages are outlined as well as its past, present, and future development towards full clinical readiness.

INTRODUCTION

The development of electromyographically controlled prosthetic limbs heralded a new era in the functionality and cosmesis of artificial limbs. Yet today, even the most advanced clinical myoelectric prostheses offer only a limited degree of control. One method of increasing limb control would be the introduction of decision making and "mathematical intelligence" into the limb controller. Typically, researchers in the area of myoelectric control use computers to supply this intelligence if it is needed. However, there are no myoelectric limb controllers currently available which could implement any of the newly developed control algorithms. Examples of these algorithms can be found in Hogan [1], Graupe [2], Shwedyk [3], and a host of others. Current limb controllers use analog circuitry which cannot supply the intelligence needed. However, the microprocessor is ideal for use in a situation which requires intelligence. The idea of using a microprocessor in a myoelectric limb controller is not new (see Graupe 1977 [4]), however, power consumption and cost have always been a limiting factor to its development. However, now the technology is available to develop clinical microprocessor based myoelectric limb controllers [5]. The following paper describes the design, implementation, and testing of such a device.

HARDWARE DESIGN

The described limb controller is created to be used as a demonstration project and design tool, not as a final end product. Therefore, in its final implementation, some sections of the present limb controller may not be necessary. The present design allows the removal of sections without necessitating a complete redesign.

The limb controller presently consists of five printed circuit boards each approximately 23 cm. (9 in.) square. Since this is only the prototype the size was purposely exaggerated to aid in the development process for it does not make sense to miniaturize a limb controller until it is ready for full clinical trials. A detailed description of the design cannot be included here, but may be found in Hortensius [5].

It was decided that four EMG signal channels should be made available to the limb controller. Using four channels provides the limb controller with much more information than the two channels typically used in current limb controllers. Also, this will allow further research into multichannel EMG signal processing within the confines of a prosthetic limb controller.

The heart of the limb controller is the "intelligence" that is provided by the microcomputer. The microcomputer has the responsibility of deciding what action should be taken based on the present EMG signal(s). For discussion purposes the limb control microcomputer is divided into three sections; data acquisition; EMG signal processing; and motor control.

The data acquisition and motor control sections use an MC146805E2 microprocessor, while the signal processing is done by a CD80C86 microprocessor. The use of two microprocessors in a system with low power and small space requirements may seem extravagant, but, in fact, the arrangement is quite efficient. The MC146805E2 is an eight bit microprocessor with sixteen input/output lines. Therefore, it can be easily interfaced to other devices. However, its data processing ability is poor because of a limited instruction set that is customized for use in control applications. As a result, the MC146805E2 becomes cumbersome and slow when doing any mathematical analysis. The CD80C86 is a sixteen bit microprocessor that has no input/output lines but can do many mathematical functions quickly and efficiently. Therefore, while the CD80C86 has good data processing abilities it requires additional circuitry to interface it to other devices. Thus, a combination of the two devices, where the MC146805E2 gathers the data and implements the control output that the CD80C86 has determined, is actually an efficient system. This also allows the controlling computer to react more quickly than if only one microprocessor were used to gather and process the data and then implement the desired control on the prosthetic limb. The two microprocessors communicate via a bank switched shared memory which allows both microprocessors to operate at full speed even during shared memory accesses. The CD80C86 operates at a clock rate of 4.77 MHz and the MC146805E2 operates at 2.38 MHz. Memory consists of 4k words of control memory and 2k words of local variable memory for the CD80C86, 4k bytes of control memory and 128 bytes of local variable memory for the MC146805E2, and 4k words of shared variable memory. The above hardware allows the limb controller to operate in real time running under most of the new myoelectric control algorithms.

A unique feature of this limb controller is the capability for limb position feedback from transducers mounted on the prosthetic limb. This closes the feedback loop thereby providing more accurate prosthetic limb control and also creating many possibilities in the area of preprogrammed multifunctional movements. A block diagram of the limb controller is given in Figure 1.

Table 1 shows the limb controller's power usage. The battery to be used with this limb controller has a capacity of 8.6 watt-hours which is sufficient to operate the limb controller for about 3 hours. This is unacceptable for clinical requirements but it should be noted that the data acquisition portion of the limb controller uses about 61% of the power. Designs using less power are available and can be incorporated into the limb controller as the need arises.

	power used
Data acquisition section	1250 mW
Motor control section	200 mW
Microcomputer section	600 mW
Total limb controller load	2050 mW

Table 1 Limb Controller Power Usage.

SOFTWARE DESIGN

The limb controller software is separated into three parts, limb control, EMG signal acquisition, and myoelectric control. This is done primarily because the hardware is also divided along these lines and secondly because this leads to an efficient software implementation. As mentioned above, the MC146805E2 is best suited to output control and data acquisition while the CD80C86 is best suited to myoelectric limb control.

The movement and data acquisition software is written in such a way that it is portable and can operate with any myoelectric limb control algorithm that may be in operation. This is done by essentially calling the myoelectric limb control algorithm on the CD80C86 as a subroutine of the MC146805E2 software. This ensures that the only software that needs to be modified during investigations into various myoelectric limb control algorithms is the actual myoelectric limb control software on the CD80C86.

Several software features provide the described limb controller with advantages over present myoelectric limb controllers in both controllability and functionality. These features include; up to 64 preprogrammed multifunctional movements, improved movement control using joint position feedback, automatic return to start position, superimposed movements, joint out of range control, and proportional speed control of joint motors. Myoelectric limb control algorithms that have already been implemented and tested on the limb controller include, two and three state control and the four coefficient autoregressive EMG signal model developed by Graupe [2]. User operation under these algorithms varied; all test subjects could easily operate the two state control and after about an hour or two of practice could operate the three state control. However, the success rate for the four coefficient autoregressive signal processing model was much lower. The main result from the autoregressive signal processing is to display the ability of the limb controller to handle a complex, calculation intensive, signal processing algorithm in a real time environment with an acceptable reaction time of 100-200 milliseconds.

Development of a Clinical controller

The research undertaken for this project involved the preliminary steps in the development of a clinical microcomputer based limb controller. The first part of the investigation involved an extensive literature search to find other projects in the myoelectric limb control area that had similar goals or objectives. This survey revealed that very few attempts had been made to incorporate microprocessor based technology into a myoelectric limb controller. As well, any attempts that been made were dated by several years. However, several promising myoelectric control algorithms were found.

The next phase of the project involved the hardware and software design of the limb controller. A first prototype was designed and constructed and it demonstrated that the principles behind this limb controller would work. However, reliability problems necessitated the construction of the present more robust controller.

Future development to bring the limb controller to a clinically usable stage should involve the study of other limb control algorithms in order to find one which can provide sufficiently improved performance to justify the costs of miniaturizing the limb controller hardware. An interesting point is that the limb controller can already provide better two and three state control than that of clinical limb controllers because of the flexibility that is provided in adapting the software to a particular user's need. Therefore, the limb controller is already at a stage where it could be readied for full clinical trials.

To ready the limb controller hardware for full clinical trials, sections of the limb controller that are not needed by the selected control algorithm should be removed. The remaining components should then be miniaturized through component hybridization and/or the use of custom designed integrated circuits. This is an expensive proposition and so should not be undertaken until the final version of the limb controller has been determined.

SUMMARY

The design, implementation, and testing of a microcomputer based myoelectric limb controller has been described. The limb controller provides the possibility of enhanced control of prosthetic limbs through the use of its microcomputer "intelligence". It presently is at a developmental stage but when ready for full clinical trials needs only ~~to~~ be miniaturized. The limb controller has demonstrated the ability to duplicate the two and three state control of current clinical myoelectric limb controllers as well operate under new advanced computationally complex myoelectric limb control algorithms.

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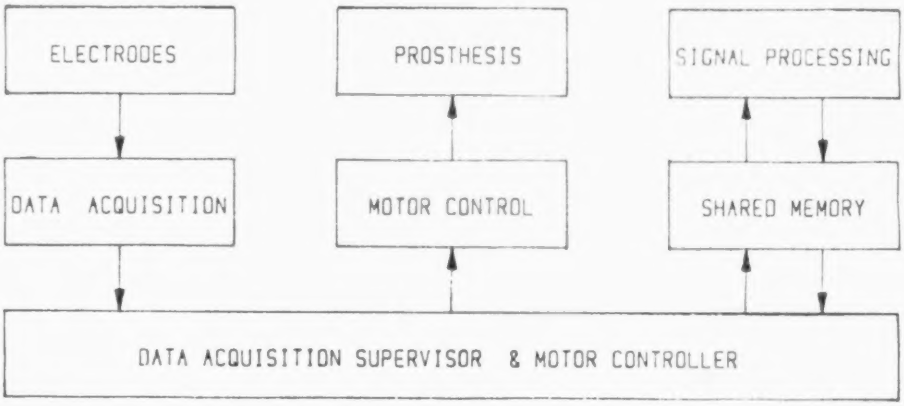

Figure 1. Block Diagram of Limb Controller.

NO MORE SPECIAL ME

Christa DeSantis
The Kids on the Block Inc.;
822 North Fairfax Street
Alexandria, VA 22314

The Film No More Special Me was produced in Perth before a live audience of disabled and non-disabled children. The purpose was to show how The Kids on the Block puppets encourage non-disabled children to ask and learn about their handicapped peers. Hosted by Bert Newton, a well-known Australian entertainer and commentator, the show features child-size puppets who describe their abilities and disabilities in a clear uncomplicated way. Renaldo is blind, Mandy is deaf, Ellen Jane is mentally retarded, and Mark has cerebral palsy and uses a wheelchair. Each puppet interacts with another non-disabled puppet character and demonstrates to the audience how handicapped and non-handicapped children can become friends.

As demonstrated in the question and answer exchange, children have a natural curiosity about disabilities. "Does it hurt to be blind?" one child asks Renaldo. "What's the hardest part about being retarded?" another asks Ellen Jane. Curiosity about disabilities is an understandable reaction and one that must be expressed if handicapped and non-handicapped children are to build positive relationships in the mainstream of society. As well as showing The Kids on the Block in performance, No More Special Me includes several segments in which educator Barbara Aiello discusses such issues as mainstreamed classes, the use of sign language, the myths and misgivings that surround many disabilities, and the great benefits to be gained in knowing disabled people.

SPINAL CORD STIMULATION IN THE TREATMENT OF MOTOR DISORDERS

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During the past 10 years we have used spinal cord stimulation in the treatment of over 650 cases with motor disorders. This therapeutic application has been expanded by us to include cerebral palsy and other related neuromuscular disorders, such as dystonia, torticollis, multiple sclerosis, spinocerebellar degeneration, Friedreich's ataxia and spinal cord injury.

An integral part of our investigational protocol is the videotaping of all patients preoperatively, post-operatively and at follow-up intervals. Examples of the therapeutic effectiveness of spinal cord stimulation in these various disorders will be shown.

The technique of percutaneous implantation of the computerized multi-level spinal cord stimulation system will be shown to demonstrate the simplicity of the procedure, which carries with it little or no risk.

Our results have indicated that the level of the spinal cord stimulated, the polarity of the electrical field and the frequency of the stimulation are critical to achieve a satisfactory result and must be individualized in each case regardless of etiology. The procedure for carrying out these analyses will be demonstrated, which can be done by the patients primary physician after the implant by the neurosurgeon.

TWENTY IS PLENTY

Trish North
Professional Education Program - United
Cerebral Palsy of New York City, Inc.
122 East 23rd Street; New York, New York 10010

"Twenty is Plenty" was written, produced and directed by Don Brockway, former director of Audio-Visual Services for United Cerebral palsy of New York City. The idea of a training tape dealing with staff treatment of the disabled was originally conceived by a client of the agency, Tony Peluso, who subsequently consulted on the script.

Using twenty amusing vignettes, the tape focusses on such issues as the difficulty in understanding some disabled persons' speech, underestimating a client's abilities, treating the adult disabled as children, among others.

PTOLEMY, ARISTOTLE AND SHARON THORNE'S FAMILY

Trish North
Professional Education Program - United
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Produced for in-service training to facilitate better relations between staff and families of clients, this tape tackles some tough issues which face rehabilittaion workers. Approached from the point of view of a concerned teacher, the issue of family accountability and cooperation with staff is explored.

Parents of a disabled child who is not doing well in programs are shown grappling with myriad problems. Hardly disinterested in the child's welfare, they are instead overwhelmed by the problems the child's condition creates for the family. Using a clever play on tenets of Greek philosophy, the tape attempts to show that assumptions by staff about a family's interest may be invalid. The tape concludes with the happy results of a newly formed relationship between staff and the family.

SPECIAL FRIENDS

Myra W. Kessel and UTREP Staff
University of Tennessee Rehabilitation
Engineering Program; 682 Court Avenue;
Memphis, Tennessee 38163

The "Special Friends" Program, formerly known as the Adopt-A-Child Program, was started in 1981 by the University of Tennessee Rehabilitation Engineering Program (UTREP). It was initiated as a result of cutbacks in funding for technical services at both the State and Federal levels.

The "Special Friends" Program is designed to help families obtain financial assistance to pay for prescribed services and to aid those who have traditionally fallen in the funding gaps. The inability to pay for services can result in a denial of services to clients and families often in the greatest need.

The "Special Friends" Program seeks to bring together those families in need with individuals, service groups, clubs and/or businesses who may be willing to help. An 8 minute slide/sound presentation was developed by the staff and several clients of the UTREP, to be used as an educational and fund raising tool. This audio visual presentation seeks to educate those who view it as to what Rehabilitation Engineering means and why some persons are in need of financial assistance.

Over the past four years, this program has received over $33,000.00 in contributions and has aided over 55 clients.

COMMUNICATION IS NOT JUST WORDS

Ross Breadner
Thames Valley Children's Centre
385 Hill Street; London
Ontario, Canada N6B 1E4

This sound/slide presentation provides indepth information about communication systems for those children who are unable to speak or have limited speaking abilities. It was developed as a teaching tape for professionals, primary care-givers, students and parents to help them to acquire a better understanding of this area.

The following topics are discussed and illustrated:

I. Factors to consider in the selection of an augmentative/alternative communication system.

II. Communication devices available, outlining special features of each device, means of accessing, prerequisites needed for their use.

III. Misconceptions about nonverbal communication systems as well as suggestions for a listener communicating with a nonverbal child.

NEW INFANT VISION SCREENING TECHNOLOGY

John Brabyn, Ph.D.
Smith Kettlewell Institute of Vision Sciences;
Medical Research Institute at Pacific; 2200
Webster St., San Francisco, California 94115

Ths narrated slide-show presentation will illustrate the development and application of a new vision screening system based on the visual evoked potential (VEP). The particular technique described, which was developed at Smith-Kettlewell, is capable of giving an acuity estimate of any subject (including a young infant) merely by requiring him to watch a video display for ten seconds while his VEP is monitored. The "fast sweep" method used makes the technique especially suitable for infants, whose attention spans are short. Accordingly, we have established a pediatric diagnostic center to examine the clinical implications of the new technique. Results from both normally and abnormally developing infants are presented, and the system's potential for use in early diagnosis and intervention in visual disorders is assessed.

REHABILITATION ENGINEERING "A CHALLENGE FOR ALL"

Terry Willkomm
Breaking New Grounds - Purdue University,
Department of Agricultural Engineering
West Lafayette, Indiana 47907

This presentation is an overview of what rehabilitation engineering is and how everyone can become involved.

Primary objective of the presentation is to show professionals and consumers that they do not need to be an engineer to solve unique problems.

Specific areas addressed include: utilizing community resource people, places and unique materials; approaches to solving unique problems; low cost, low technology devices (over 46 slides of specific examples); accessing commercially available devices and fabricating materials.

FEEDING

Ross Breadner
Thames Valley Children's Centre
385 Hill Street; London,
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This sound/slide presentation is an introductory teaching tape for professionals, primary care-givers, students and parents involved with children with feeding problems due to neuromuscular disorders. The relationship among feeding, speech and communication is empahsized throughout the presentation, as well as the need for an interdisciplinary approach when developing an intervention program.

The presentation provides an overview of the following:

I. Normal developmental stages of feeding and progression of food types.

II. Feeding problems and interventions to improve/aid feeding skills. Some specific types of interventions in areas of environment, positioning, food types and adaptive equipment are illustrated.

REHABILITATION AIDS: POSITIONING BELTS AND PRONE-SUPPORT WALKER ("THE HORSE")

Terrand B. Grall
Consumer Care Products/Consulting Inc.
6405 Paradise Lane
Sheboygan Falls, Wisconsin 53085

This 25-minute videotape presentation includes a detailed descriptive and demonstration of two ergonomically designed rehabilitation aids, i.e., Positioning Belts and the Prone-Support Walker, also known as "The Horse."

The unique design of these products attributes to several advantageous characteristics that will be noted throughout the entire presentation. For example, the special features of each aid allow for individual fit adjustment, ease of use (requiring few, if any, tools), comfort, and versatility (providing more than one use/application).

The presentation is divided in two parts, the first part focusing on the assembly, components, and special features of Positioning Belts and a demonstration of their application to a wheelchair. Part two provides a demonstration of the Prone-Support Walker's use with two handicapped students and an interview with a physical therapist and special education classroom teacher, with an emphasis on the therapeutic goals enhanced through its use.

CURRENT MYOPROSTHETIC DEVELOPMENTS AT THE HUGH MACMILLAN MEDICAL CENTRE

M. Mifsud, Galway, and M. Milner
Hugh MacMillan Medical Centre - Rehabilitation
Engineering Department; 350 Rumsey Road;
Toronto, Ontario, Canada M4G 1R8

Four new devices developed to augment the fitting of myoelectrically operated upper-extremity prostheses are presented.

1) A miniature data acquisition device (DAD) to record both the number of open and close events and the cumulative event duration has been developed. DAD can be placed within the prosthesis to objectively collect information regarding prosthesis use during the amputee's normal daily activities.

2) A self-adaptive myoelectric processor (SAMP) is a 3-state system developed to: facilitate the fitting and training process; optimize the calibration settings as required; motivate prosthesis use through immediate operation; and minimize health care cost and disrupted schedules for recalibration purposes.

3) A microcomputer-aided myoelectric control assessment program (MCAP) is being developed for routine clinical use to objectively measure, in the clinic, amputee responses to a visual image while operating their prescribed myoelectric prosthesis.

4) A microcomputer-based device has been developed to augment the process of identifying suitable muscle control sites. Graphical and numeric display of the myoelectric voltage signal levels from up to 2 separate muscle groups is complemented with the display of previous results obtained while mapping the muscle bulk area.

ONLY YOU CAN PREVENT PRESSURE SORES

Kathleen J. Manella
Orthopaedic Hospital
2400 South Flower Street
Los Angeles, California 90007

The presentation focuses on the hygiene, nutrition, and pressure-relief mechanisms as they relate to pressure sore prevention. The presentation is specifically designed for the non-clinical patient and for the clinician as a teaching tool to foster better patient cooperation.

HEALTH ASPECTS OF FOSTER CARE

David P. Schor
University of Iowa; Department of Pediatrics;
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Iowa City, Iowa 52242

This videotape and accompanying monograph will provide information to foster care agencies, workers, and parents enabling them to become better aware of the health care needs of children in foster care. The material will assist these professionals in their efforts to improve the health status of foster care children.

The videotape profiles three children newly placed in foster care, describing the events precipitating their placements and the impact of those placements on the children. The viewer is provided techniques for easing the transition: use of a "lifebook" scrapbook, for example. A social worker demonstrates how the children's initial health care needs were identified and met. The program presents guidelines for routine health care supervision and methods to meet the special care needs of foster children. Finally, the viewer learns the outcome for the profiled children.

The accompanying monograph details the special medical and psychological needs of foster care children as well as stressing their need for routine well-child care. It empahsizes the importance of timely and thorough record-keeping by both foster care agencies and foster parents and notes two exemplary record forms. It defines and describes "screening tests", anticipatory guidance," and other terms. It recommends a two-part "health status check" as part of each foster care placement. Appendices list some special medical needs of children in foster care, provide a suggested schedule and content of regular health supervision visits, and present recommendations for routine immunizations as well as a schedule for protecting the previously under-immunized child. A reference list concludes the monograph.

SEATING FOR CEREBRAL PALSIED CHILDREN

Douglas A. Hobson and UTREP Staff
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The advantage of adaptive seating as an adjunct to the overall management of the child or young adult with cerebral palsy is now being recognized by many therapists and other health professionals. Adpative seating, among others, places the child in a position which allows the child to utilize and maximize motor skills.

This presentation provides information necessary to evaluate and prescribe an appropriate seating system for a child with cerebral palsy.

PROJECT TEACH

Memphis City School and University of
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TEACH stands for Technical Education Aids for Children with Handicaps. It represents a 3 year early childhood demonstration project of the Federal Office of Education in which 10 children were provided with assistive devices including adaptive seating systems, mobility aids, feeding devices, and augmentative communication aids. This presentation summarizes the effects of these devices on academic performance, motor, daily living, and communication skills. The presentation also highlights the importance of the team approach to solving the technical needs of children with handicaps.

PROSTHETIC TERMINAL DEVICE FOR PLAYING THE PIANO

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A terminal device is being developed to enable a person fitted with a below-elbow prosthesis to play the piano. The device which consists of two fingers and a pivoting wrist, is activated by a foot control mechanism.

The purpose of this project is to design a prosthetic terminal device for a 10 year old girl, R.M., who indicated a desire to play the piano. The child has a congenital deficiency of the left arm, a disability equivalent to a short below-elbow amputation. She currently has a conventional hook prosthesis with a quick release wrist unit and a Figure-8 type harness.

Functional design criteria for a terminal device have been developed in conjunction with R.M.'s piano instructor. The primary requirements are based on the accompanying role played by the left hand and include:
1) variable spread of the two "fingers" to provide combinations of a third to a full octave; and 2) wrist rotation for the black and white key combinations and full range on the lower half of the keyboard.

The project objectives are to design a terminal device so that it:

1) fulfills the motion requirements,
2) can be accurately controlled using cables,
3) provides positioning feedback,
4) is light weight,
5) can be built utilizing commercial parts,
6) is esthetically pleasing.

BARRIER FREE: AN INTERVIEW WITH PAULA HAYNES

Fran Theos and Michael Heinrich
180 Biking
Cordova, Tennessee 38018

Paula Haynes, a resident of Memphis, Tennessee, tells of her experiences growing up with a congenital disability, and attending high school, as well as subsequent difficulties in finding full time employment.

With both humur and persistence she developed personal ties which led to satisfactory full time employment at Graceland. She tells of her travels to the Los Angeles area, and shares several stories relating to a series of trips with her friends to the Beach Boys Concerts.

The authors wish to acknowledge the use of the resources of Public Access Cablevision (Memphis Channel 7), their staff and volunteers.

BARRIER FREE: AN INTERVIEW WITH MARTY AND JANE DAVIS

Fran Theos and Michael Heinrich
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Marty and Jane Davis are a young married couple, both currently students at Memphis State University. They tell of the change in their lifestyle when Marty experienced a spinal cord injury in a bicycle accident. Marty, now a "strong" paraplegic secondary to this accident, has continued his active lifestyle. Marty and Jane discuss their summer travel plans which they see as the last opportunity for a travel fling prior to settling down to the work-a-day world, since their graduation from Memphis State University is scheduled for December 1985.

The authors wish to acknowledge the use of the resources of Public Access Cablevision (Memphis Channel 7), their staff and volunteers.

BARRIER FREE: AN INTERVIEW WITH DR. BILL DELOACH

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Dr. Bill Deloach is a professor at Memphis State University. Dr. Deloach tells of being injured in a diving accident while a sophomore in college. He relays his experiences of completing his delayed college education and successful job search while recovering and adjusting to this high level spinal cord injury. In a frank and open manner he discusses some of the difficulties of travel as a wheelchair user and also provides a series of practical and informative helpful hints for the disabled traveler.

The authors wish to acknowledge the use of the resources of Public Access Cablevision (Memphis Channel 7), their staff, and volunteers.

NEUROBEHAVIORAL MATURITY ASSESSMENT FOR PRETERM INFANTS - A NEW APPROACH

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Thomas Forrest
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Development of our Neurobehavioral Maturity Assessment began in 1977 in the context of a intervention study with very young preterm infants. We needed a fine-grained instrument to monitor in weekly intervals the developmental progress of an experimental and a control group.

The primary purpose of our Procedure is to test the maturity of the infants' functioning. The scores of each scale are therefore ordered from the least to the most mature responses. The examination can also alert the clinician to potential neurological abnormalities. Additionally, stable individual differences can be detected by examining infants over time.

The Procedure has three new features:

1. It has a cumulative structure which permits evaluation of the neurobehavioral functioning of an unusually wide range of preterm infants (from 25 weeks conceptional age to term). Part I is a very short procedure which assesses with a minimum of handling the tone, motor patterns and the states of the infants. Because this part of the examination takes no more than 10 to 12 minutes and relies heavily on observational items, it has been used with very young, intubated perterms who were sufficiently stable medically to be examined. Part II, which starts after part I is completed, consists of scales which test more mature forms of behavioral responses. Part II can be administered to medically stable infants who are on room air, starting at 32 weeks conceptional age.
2. A truly invariant sequence for administering the test items is used. This standard sequence is designed to bring about those behavioral states which are likely to elicit the best possible responses from preterm infants. By building into the Procedure a standard sequence of rousing, soothing, and alerting items, the opportunity to test various functions in appropriate behavioral states is maximized and the need to intervene with some infants more than with others is minimized. This prevents the examination from becoming a different experience for each child.
3. For analyzing the longitudinal data, we developed a mathematical model which permits an assessment of how gestational, conceptional and chronological ages and birthweights impact on the development of different functions and how "catch-up" time of performance differs according to function.

This work is supported by Grant MH 36884 from the National Institute of Mental Health Center for Prevention Research, Division of Prevention and Special Mental Health Programs and by Grant RR-81 from the General Clinical Research Centers Program of the Division of Human Resources, National Institutes of Health.

A CRY FOR HELP: THE PLIGHT OF THE SEVERELY HANDICAPPED CHILD AND HIS FAMILY

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Yesterday's medical "miracles" have become commonplace technology in the newborn nursery and the pediatric intensive care units and are enabling a higher percentage of fragile newborns and severely brain injured children to survive. Some of these children will carry with them permanent and significant medical disabilities that often require around-the-clock care. This vigilant care must often be provided by the families of these children who have no medical training and limited supportive medical services available except for acute crisis intervention.

The burden of these families is a heavy one. The medical, social, and psychological stresses often seem insurmountable. There is no time for routine activities without medical responsibilities intervening; other family responsibilities are neglected; family members become alienated to the point of division.

This documentary was developed by a parents action group in New Mexico in an attempt to bring this situation to the attention of the public. The families portrayed here are typical examples of the many such families that exist in every state, each in need of better and more available help from several areas. There is presently legislation pending in the State Government of New Mexico to make it the first state to have a comprehensive, well funded support system available to these children and their families. the proposed program would include in-home nursing care, adequate family training and counseling, out-of-home respite care, companion home care setting, and small pediatric nursing homes. Cooperation between State and Federal agencies must be obtained. A special Medicaid Waiver is currently being sought to make the in-home care and other options an accessible reality for these families.

Society has again found itself out of step with the results of medical and technological gains, and must now take some major responsible steps in an attempt to catch up.

MANAGEMENT OF THE CHILD FED VIA GASTROSTOMY TUBE: AN ORAL FEEDING APPROACH

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The purpose of this audiovisual presentation is to acquaint health care professionals with the steps necessary to develop oral feeding skills in children who have been nourished by gastrostomy tube for long periods.

Children nourished by gastrostomy tube can pose difficult management problems for both families and professionals. Lack of progres in oral feeding, as well as the threat of poor weight gain, often make families give up their attempt to reinstate oral feedings. To help families help their children develop oral feeding skills, a systematic interdisciplinary approach is advocated.

This self-instructional videotape illustrates interdisciplinary methods for overcoming the medical, physiological, nutritional and behavioral barriers that interfere with oral feeding. The program includes strategies for determining oral feeding candidates, guidelines for medical and nutrition preparation, oral feeding techniques and guidelines for gastrostomy tube removal. This program is illustrated by following a child, his family and professional staff through the evaluation and treatment process.

At the conclusion of the video presentation, the viewer will have knowledge of the principles and methods described. The experienced health care professional will have sufficient information to apply the program in their own clinical settings.

LAINEY

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There are thousands of school age children in the United States who are being excluded as active participants in their educational process. These individuals whose intellectual abilities are within or above normal levels, but who are so severely physically handicapped that they cannot voluntarily use their body parts to speak, write, manipulate objects, sit or move independently.

For more severely involved, the problem resides in their inability to access available technical aids, even with single switches. These students often have only their head control to call on for function. This one body part must be able to operate a powered mobility device, a communication aid, and also the input to a computer system. Any control system developed must be portable and have multiple functions if it is going to maximize the very limited motor skills.

This videotape presents the results of a cooperative effort between the Prentke Romich Company with a subcontract to the University of Tennessee Rehabilitation Engineering Program. The system developed is a modular component system which provides appropriate mobility, seating, control and interface with a communication aid and computer. With only one input mode required the system allows the severely disabled to compete on a more equal footing in a broad range of life situations. More important, it depicts the impact of technology on a young lady and her family and how she has been able to obtain a degree of independence.

HOW MOTOR DEPRIVATION AFFECTS LEARNING

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This program deals with the role of normal motor development in the newborn and how motor deprivation affects the acquisition of conceptual, perceptual and behavioral skills. The differences between motor behaviors in the normal and brain damaged child is presented as well as the effects of abnormal reflexes. The effect of normal motor learning on learning in other areas is considered.

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